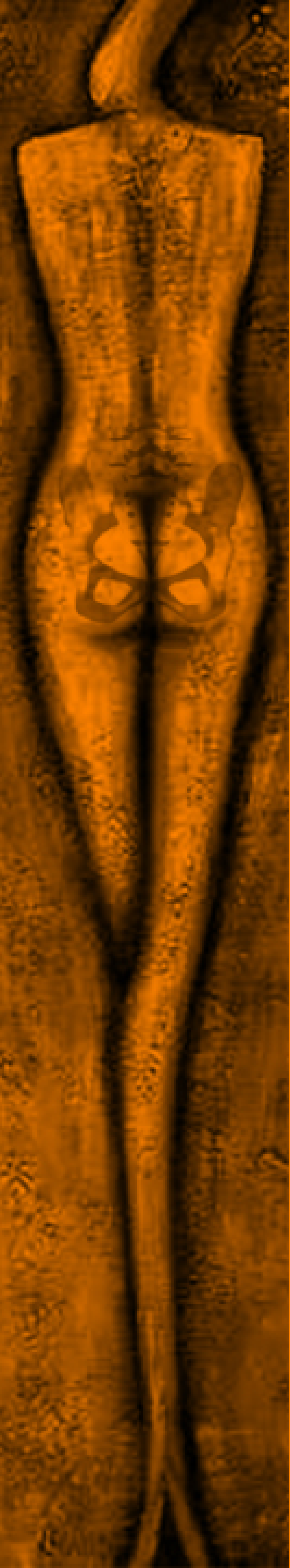
Motor Control and Lumbopelvic Stability in Young Healthy Women

Hu, H.

2011



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献给我最亲爱的儿子胡瑞霖

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Nederlands
Orthopedisch
Research en
Educatie
Fonds

ISBN: 978-90-5335-448-3

VRIJE UNIVERSITEIT

Motor Control and Lumbopelvic Stability in Young Healthy Women

ACADEMISCH PROEFSCHRIFT

ter verkrijging van de graad Doctor aan
de Vrije Universiteit Amsterdam,
op gezag van de rector magnificus
prof.dr. L.M. Bouter,
in het openbaar te verdedigen
ten overstaan van de promotiecommissie
van de faculteit der Bewegingswetenschappen
op maandag 26 september 2011 om 13.45 uur
in de aula van de universiteit,
De Boelelaan 1105

door

Hu Hai

geboren te Fujian, China

promotoren:	prof.dr. J.H. van Dieën
	prof.dr. B.J. van Royen
copromotor:	dr. O.G. Meijer

Contents

Chapter 1

General introduction

Background

The impetus for this thesis stems from an ongoing series of studies on pelvic girdle pain (PGP, Wu et al., 2004a; Vleeming et al., 2008). PGP is regarded as a specific form of low back pain (LBP) that can occur separately or in conjunction with LBP, as quoted in the European Guidelines (Vleeming et al., 2008):

> *PGP is defined by pain experienced between the posterior iliac crest and the gluteal fold, particularly in the vicinity of the sacroiliac joints (SIJ). The pain may radiate in the posterior thigh and can also occur in conjunction with/or separately in the symphysis. The endurance capacity for standing, walking, and sitting is diminished. PGP generally arises in relation to pregnancy, trauma or reactive arthritis.*

In a systematic review, Wu et al. (2004a) observed that the far majority of PGP cases are related to pregnancy, with complaints starting during or shortly after pregnancy. Wu et al. (2004a) coined this syndrome “pregnancy-related pelvic girdle pain (PPP)”, and reported that about 45% of all pregnant women and 25% of all postpartum women suffered from PPP and/or pregnancy-related low back pain (Wu et al., 2004a). A wide number of health-related specialties, including obstetrics and gynecology, general medicine, orthopedic surgery, physiotherapy, rheumatology, and psychiatry have been involved with PGP, both scientifically and clinically.

Pain is the most obvious symptom in PGP, which can be stabbing, dull, shooting, or burning, and is mostly rated at around 50 to 60 mm on a visual analogue scale. Pain localization, onset and intensity vary. Importantly, PGP causes disability, usually mild, but severe in about 8% of cases (Wu et al., 2004a). In most instances of PGP, motor functioning is affected with considerable impact on activities of daily life, and patients often experience problems with walking, standing and sitting (e.g., MacLennan and MacLennan, 1997). Many patients cannot walk quickly or cover long distances (Fast et al., 1987; Hansen et al., 1999; Mens et al., 1996). Moreover, during walking, women with PGP reported a sensation as if hip flexion was temporarily blocked (a ‘‘catching’’ sensation, Sturesson et al., 1997). This may be related to the feeling ‘‘as if the leg is paralyzed’’ during the Active Straight Leg Raise test, in which one leg is raised by the patient in a supine position (Mens et al., 1999). These unusual findings appear to suggest problems in the control and/or coordination of walking, more specifically of hip flexion in walking. Alteration of the gait pattern has also been associated with PGP. Gait coordination in these patients is characterized by a lower walking speed, an increase in the amplitude of the horizontal rotation of the pelvis and a reduced relative phase between rotations of pelvis and thorax. The increased pelvis amplitude may differentiate PGP patients from those with LBP and healthy pregnant women (Van Wingerden et al., 2008; Wu et al., 2004b, 2008).

A thorough medical history, together with physical examination, and appropriate laboratory tests to exclude other causes of symptoms, should underlie the diagnosis of PGP. Validated clinical tests include pain provocation tests (P4/thigh thrust, Patrick's Faber, Gaenslen's test, and the modified Trendelenburg's test) and pain palpation tests (of the long posterior SI ligament test and the symphysis). Furthermore, as a functional test, the active straight leg raise (ASLR) test is recommended (Vleeming et al., 2008). Because most of these tests have a reasonable specificity but low sensitivity, there appears to be a consensus for a combined use of all of these tests to minimize false-negative results (e.g., Vleeming et al., 2002). The ASLR is often used as a preliminary test (e.g., Ronchetti I, 2008), and is assumed to assess load transfer between the trunk and the lower limbs (Mens et al., 1999).

Unfortunately, the pathophysiological mechanisms underlying PGP remain poorly understood. A variety of etiologic factors have been proposed, such as hormonal and biomechanical factors associated with pregnancy, and problems with motor control (O'Sullivan and Beales, 2007; Kanakaris et al., 2011). Intervention studies report mixed success, with a lack of overall effects (Nilsson-Wikmar, 2005; Haugland, 2006; Östgaard, 1994). Clearly, underlying mechanisms have to be understood better before this situation can improve. Recent literature focuses on the mechanisms that play a role in the ASLR.

General aims of the study

Pelvic stability in the ASLR

In the Netherlands, PGP is generally known as "pelvic instability" (bekken-instabiliteit). The term is also commonly used in Australia ("the Pelvic Instability Association"), but in fact, it remains unclear whether instability of the pelvis plays a causal role. According to Snijders' group, stability of the SI joints is dependent on two mechanisms (Vleeming et al., 1990a & b): "form closureò and "force closure". Form closure depends on the irregularity of the articular surface of the sacroiliac joints, and the wedge shape of the sacrum. In addition, stiffness of ligaments produces forces to counteract shear loading across the SI joint (Snijders et al., 1993a & b; Vleeming at al., 1990a & b, 1996).

It is widely accepted that the SJ joints, together with the supporting ligamentous structures, may be a source of pain (cf. O'Sullivan et al., 2002). Some biomechanical models relate low back pain to an overloading of the SI joints and nearby ligaments, such as the iliolumbar ligaments (Pool-Goudzwaard et al., 2001). Several tests, such as the thumb-posterior superior iliac spines test, the heel-bank test, and the abduction test, have been developed to pinpoint sacroiliac problems in PGP, but have a poor intertester reliability (Van Kessel-Cobelens et al., 2008). By means of Doppler imaging of vibrations, a significant relation was observed between asymmetric laxity of the sacroiliac joints and pelvic girdle pain during pregnancy (Damen et al., 2002).

Changes in SI joint properties cannot account for all cases of PGP. In fact, the passive structures provide insufficient stability to the pelvic girdle. Ligaments cannot be used continuously, because creep would render them unsuitable to maintain stability (Richardson et al., 2002; Pel et al., 2008). Nearly two decades ago, Panjabi (1992) proposed a model of spinal stability, relying on three subsystems: the passive structures of the spinal column, active spinal muscles, and the neural control system. Recent research has focused on alterations in motor control as potential mechanisms underlying LBP. In LBP, Paul Hodges and his group (e.g., Hodges and Richardson, 1996) reported that the deep trunk muscles are usually activated later and are less active than normal in anticipating perturbations. These findings provided the basis for novel exercise interventions of spinal pain (e.g., Tsao and Hodges, 2008).

Similar as in Panjabi's model for spinal stability, pelvis stability would not only depend on the joint properties but also on the muscles surrounding the pelvis and their control. Evidence for the effectiveness of a motor learning approach in the management of PGP supports that motor control deficits may underlie some of these disorders (O'Sullivan and Beales, 2007; Stuge et al., 2004). Therefore, disturbed motor control may contribute to instability of the pelvis. As stated above, patients with PGP appear to have problems with hip flexion in walking (Sturesson et al., 1997), and in the ASLR (Mens et al., 1999). Anatomy textbooks present a large number of muscles as potential hip flexors, including iliacus, psoas, rectus femoris, sartorius, tensor fasciae latae, pectineus, adductor longus, and gracilis. Most of these muscles insert on the ilium and femur or in case of the rectus femoris on the tibia, while the psoas inserts on the ventral aspect of the lumbar spine and the femur.

Hip flexors connecting to the ilium, such as the iliacus and rectus femoris, may cause a relative forward rotation of the ilium, and thereby stress structures such as the long dorsal sacroiliac ligament, which has been implicated in PGP (Vleeming et al., 1996). Thus, disproportional activity of hip flexors that are connected to the ilium may provoke PGP. On the other hand, the abdominal muscles are considered important contributors to pelvic stability (Snijders et al., 1995; Richardson et al., 2002). Contraction of the transverse abdominal muscles may press the iliac bones against the sacrum while stressing the interosseus ligaments (Figure 1). This is Snijders' second mechanism for pelvic stability, i.e.,"force closure" (Vleeming et al.,1990a & b; Snijders at al., 1993a & b). The possible role of hip flexors in destabilizing the pelvis has remained insufficiently clear. More important, the evidence behind force closure is still insufficiently strong. Our first study, **Chapter 2**, charted muscle activity during the ASLR in healthy subjects, to analyze the potential destabilizing effects of hip flexors, and to test the hypothesis that the abdominal muscles play a role in stabilization of the pelvis during hip flexion.

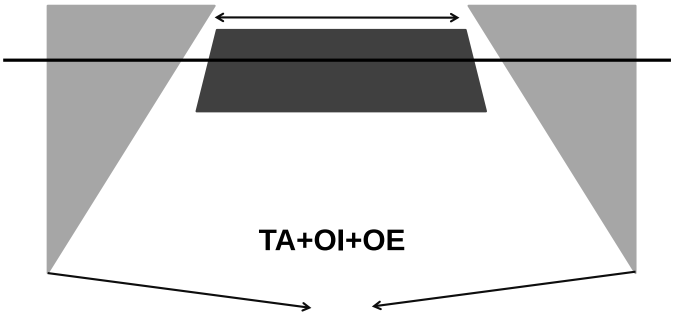

Fig. 1 Force closure by the lateral abdominal muscles

The implicit argument in the PGP literature (e.g., Mens et al., 2000) is that force closure may be affected because of problems with the deep abdominal muscles. Based on this line of argument, a pelvic belt, an elastic strap that is placed around the waist over the superior iliac spines, is seen as a mechanical device to replace the function of transverse abdominal muscles. Pelvic belts have been used in PGP for centuries (e.g., Cederschjöld, 1839, cited in Genell, 1948). Patients generally report to be satisfied with it (Östgaard et al., 1994), but randomized clinical trials so far failed to find any significant effect (Östgaard et al., 1994; Depledge et al., 2005). In fact, it has been reported that a pelvic belt does lead to stabilization of the SI joints (Vleeming et al., 1992), and to better results of the Active Straight Leg Raise in PGP patients (Mens et al., 2006b). Our study of the ASLR therefore included a condition with a pelvic belt. Moreover, since problems with hip flexion have been reported both in the ASLR and in walking (Sturesson et al., 1997), subjects were also invited to walk.

Lumbar spine stability in the ASLR

From the work of Panjabi (1992), we know that muscle activity is required to stabilize the lumbar spine (cf. Van Dieën et al., 2003). During the ASLR, sagittal plane stability of the lumbar spine may be dealt with by co-contraction of the abdominal muscles and the erector spinae (Cholewicki et al., 1997). More important, however, the ASLR is essentially a unilateral task, in which more ipsilateral than contralateral muscle activity has to be expected. Thus, the ASLR appears to offer a major challenge to frontal plane stability of the lumbar spine. While bilateral activity of any trunk muscle (Cholewicki and Van Vliet, 2002) could contribute to lumbar stabilization, given its anatomical position, the m. psoas is an obvious candidate to be involved in frontal plane stability of the lumbar spine during the ASLR.

Anatomical textbooks, such as Gray's Anatomy (Standring, 2008) emphasize that the psoas has virtually the same insertion on the trochanter minor of the femur as the iliacus. Together, these muscles are said to form the "iliopsoas", with hip flexion as its most important function (cf. Bogduk et al., 1992). Still, the psoas originates from the lumbar spine (Bogduk et al., 1992), rather than the ilium. Because of the deep localization of the psoas, it has received relatively little attention in the empirical literature, as it requires an invasive method, such as fine wire EMG (Andersson et al., 1995), with relatively long needles, which may appear to be scary to the subjects.

In **Chapter 3**, we investigated if the psoas plays a role in frontal plane stabilization of the lumbar spine during the ASLR. As (indirect) evidence for such a role, we took the lack of a statistical effect of side (right or left ASLR) to suggest a stabilizing role, since a role in hip flexion will require clearly unilateral activity. To increase the challenge to frontal plane stability of the lumbar spine, we used a condition with weight added to the leg above the ankle.

Multitasking in abdominal muscles

The human body comprises an intricate 3-D framework of bones, joints, muscles and ligaments, all responsible for posture and movement. The whole system needs to be controlled by the nervous system in order to produce coordinated movements and skilled actions. For instance, in force closure (Snijders et al., 1993a & b) the oblique and transverse abdominal muscles not only need to be properly coordinated among themselves, but also with other ongoing activities. Even in tasks where the constraints appear to be relatively simple, coordination often remains unclear, particularly since one may need to confront tasks simultaneously, and conflicting constraints can probably not be avoided. From this point of view, even a "simple" task such as walking is actually surprisingly complex, as reveals itself when disorders such as PGP disrupt motor function.

In **Chapter 4** we focus on multitasking of the abdominal muscles during walking. Main questions are how the motor control system deals with the many dimensions of multitasking, including stabilization of the pelvis and of the lumbar spine. Moreover, in dealing with any task, the control system has to solve the problem of conflicting constraints, an aspect of motor control that, so far, did not receive much attention in the literature. Note that the study of Chapter 4 focused on walking, while multitasking and dealing with conflicting constraints are ubiquitous challenges for the motor control system, not only in walking.

In-depth understanding of the ASLR?

Now that the role of force closure in the ASLR has been clarified (Chapter 2), as well as the importance of frontal plane lumbar spine stability during the ASLR (Chapter 3), and now that we have reached a general understanding of multitasking, and of dealing with conflicting constraints (Chapter 4), one might expect that an in-depth understanding of the ASLR would be relatively straightforward. Still, this is not so. In **Chapter 5**, an attempt is made to see which problems remain in reaching such an in-depth understanding.

Recently, several studies on muscle activity in the ASLR have been published, but results are still not in full agreement with each other, and some findings are puzzling (e.g., De Groot et al., 2008; Beales et al., 2009a & b, 2010a & b; Hu et al., 2010). We hypothesize that this state of affairs is mainly due to two different factors. First, stabilization may require symmetrical muscle activity, but if muscles are engaged in multitasking, while some of these tasks are unilateral, the presence of symmetrical task components does not exclude involvement in unilateral, that is, asymmetrical tasks. Second, the human motor system is intrinsically variable, and the ASLR is not very much constrained, allowing subjects to vary the details of its performance. In this last study, we not only used a “normal” condition, but also one with a pelvic belt, and one with weight added above the ankle. Our focus was on symmetry, and on variability.

Outline of this thesis

All data were based on measurement of healthy nulligravidae. 17 subjects were enrolled, but in some studies, the removal of bad data led to the use of less subjects. Fine-wire EMG was used to record activity of the mm. psoas, iliacus, and transversus abdominis (TA), while obliquus internus (OI) and externus abdominis (OE), rectus abdominis (RA), erector spinae (ES), rectus femoris (RF), adductor longus (AL), biceps femoris (BF), and gluteus maximus (GM) were recorded with surface EMG. Only right-sided muscles were recorded. During ASLR, the movements of the leg were recorded with LED markers and a 2×3 camera system. During treadmill walking, more LED markers were attached, on the heels, thorax (Th6), upper arms, forearms and head; the Optotrak system was calibrated and cluster markers were related to anatomical landmark.

Chapter 2 is a published paper, in which we assessed muscle activity during ASLR and walking on a treadmill at increasing speeds, and how muscle activity changes with a pelvic belt, aiming at a better understanding of normal muscle activity during ASLR. Moreover, we used a pelvic belt to validate Snijders' theory of "force closure". We assumed that the transverse and oblique abdominal muscles should be active during normal ASLR and walking, but less so with a pelvic belt.

In Chapter 3, also a published paper, we assessed psoas function in the ASLR. The literature on psoas function is insufficiently consistent, and we tried to assess if during hip flexion the psoas always has the same function as the iliacus, and if the psoas affects the hip more than the lumbar spine. By comparing the function of psoas to that of iliacus and the other hip flexors, we intended to differentiate psoas and iliacus, and pinpoint psoas function on the lumbar spine.

In Chapter 4, which is under revision for publication, we focus on control of lateral abdominal muscles during walking. TA, OI, and OE are involved in multiple functions: breathing, control of trunk orientation, and stabilization of the pelvis and spine. How these functions are coordinated has received limited attention. We studied EMG activity of these muscles and 3-dimensional moments during treadmill walking at six different speeds (1.4-5.4 km/h). Analysis methods such as PCA, and Fourier analysis were employed to analyze how the multiple functions of the lateral abdominal muscles are coordinated by the motor control system during gait, if these muscles engage in multitasking during walking, and how the control system co-activates muscles to deal with conflicting constraints.

In chapter 5, having been submitted, we aim to understand symmetric and asymmetric task components of abdominal muscle activity during the ASLR. Right TA, OI, OE, RA, RF and BF were investigated. Alternate left and right ASLR were performed in a normal condition, with pelvic belt and with a weight added just above ankle. Moreover, this study focused on the variability of task performance, which turned out to be considerable.

Chapter 2

Muscle activity during the active straight leg raise (ASLR), and the effects of a pelvic belt on the ASLR and on treadmill walking

Hu H, Meijer OG, Van Dieen JH, Hodges PW, Bruijn SM, Strijers RL, Nanayakkara PW, Van Royen BJ, Wu W, Xia C. Journal of Biomechanics, 2010;43(3):532-9.

Abstract

Women with pregnancy-related pelvic girdle pain (PPP), or athletes with groin pain, may have trouble with the active straight leg raise (ASLR), for which a pelvic belt can be beneficial. How the problems emerge, or how the belt works, remains insufficiently understood. We assessed muscle activity during ASLR, and how it changes with a pelvic belt. Healthy nulligravidae (N=17) performed the ASLR, and walked on a treadmill at increasing speeds, without and with a belt. Fine-wire electromyography (EMG) was used to record activity of the mm. psoas, iliacus and transversus abdominis, while other hip and trunk muscles were recorded with surface EMG. In ASLR, all muscles were active. In both tasks, transverse and oblique abdominal muscles were less active with the belt. In ASLR, there was more activity of the contralateral m. biceps femoris, and in treadmill walking of the m. gluteus maximus in conditions with a belt. For our interpretation, we take our starting point in the fact that hip flexors exert a forward rotating torque on the ilium. Apparently, the abdominal wall was active to prevent such forward rotation. If transverse and oblique abdominal muscles press the ilia against the sacrum (Snijders' ''force closure''), the pelvis may move as one unit in the sagittal plane, and also contralateral hip extensor activity will stabilize the ipsilateral ilium. The fact that transverse and oblique abdominal muscles were less active in conditions with a pelvic belt suggests that the belt provides such ''force closure'', thus confirming Snijders' theory.

Introduction

Puzzling problems with hip flexion are sometimes encountered in locomotor pathology. A considerable number of women with Pregnancy-related Pelvic girdle Pain (PPP; Wu et al., 2002, 2004a & 2008; Meijer et al., 2006), or athletes with groin pain (Mens et al., 2006a; Verrall et al., 2005), are limited in raising a leg from a supine position (Active Straight Leg Raise, ASLR). In ASLR, some patients felt "as though they were paralyzed" (Mens et al., 2002), and in walking, a "catching" feeling was reported (Sturesson et al., 1997).

In a large majority of cases, a pelvic belt relieved hip flexion problems in PPP (Mens et al., 1999) and groin pain (Mens et al., 2006a; Jansen et al., 2008). The belt may (cf. Snijders et al., 1998) provide sacroiliac "force closure" (Vleeming et al., 1990a & b), as claimed to be normally delivered by transverse and oblique abdominal muscles, which may be activated too late in low back pain (Hodges and Richardson, 1999), sacroiliac pain (Hungerford et al., 2003), and groin pain (Cowan et al., 2004). Nevertheless, a randomized clinical trial (RCT) of training the oblique abdominal muscles in PPP failed to have clinical effects (Mens et al., 2000). Doppler Imaging of Vibrations (DIV) suggested that a pelvic belt stiffens the sacroiliac joint (Damen et al., 2002), but the validity of this technique remains insufficiently clear (de Groot et al., 2004). More important, in an RCT of PPP treatment that included a pelvic belt (Östgaard et al., 1994), most patients were satisfied with the belt, but some got worse, and no objective overall effect could be established. Other RCTs on treatment forms of PPP that included a pelvic belt also found no overall effects of the belt (Nilsson-Wikmar et al., 2005; Haugland et al., 2006).

In a recent study of ASLR in women with PPP (de Groot et al., 2008), muscle activity turned out to be differentially affected. Bilateral surface EMG was recorded of the mm. rectus femoris, adductor longus, iliopsoas (described as "psoas major", but the electrodes were above the insertion in the groin), and obliquus abdominis externus. Patients reported more effort to perform ASLR than healthy controls, and used a higher percentage of the maximum voluntary contraction (MVC) of the m. rectus femoris ipsilaterally, and bilaterally of the mm. iliopsoas and obliquus abdominis externus. Less hip flexion force was delivered, which the authors interpreted as the women needing "more muscle force to reach the same goal" (de Groot et al., 2008, p. 72), but a much simpler explanation would be that the women with PPP had a lower MVC of several muscles, possibly because of reflex inhibition (Hurley and Newham, 1993), or some other nervous system strategy to avoid pain. In the PPP study of de Groot et al. (2008), activity of the m. adductor longus remained unaltered, whereas in groin pain one would expect involvement of the adductor muscles (Verrall et al., 2005). Hence, it may be important to differentiate which muscles are, and which are not, involved in individual patients with problems in performing ASLR.

It remains insufficiently understood, exactly which muscles are involved in ASLR, and even among healthy subjects, strategies differ (Beales et al., 2009a). The present study aims at a better understanding of normal muscle activity during ASLR. Contrary to earlier studies, both the m. iliacus and the m. psoas were recorded, as well as all muscles of the abdominal wall. Moreover, we used a pelvic belt to validate Snijders' theory of "force closure", that is, the idea that transverse and oblique abdominal muscles press the ilia against the sacrum. If the pelvic belt indeed substitutes forces from these muscles, and if one assumes that the nervous system responds accordingly, transverse and oblique abdominal muscles should be active during normal ASLR, but less so with a pelvic belt. To extend our validation of Snijders' theory into the realm of everyday motor behavior, we added a treadmill-walking task, again with and without pelvic belt.

Methods

Subjects

Participants were healthy 20-40 years-old nulligravidae, who were recruited by word of mouth. Exclusion criteria were: previous orthopaedic surgery, walking-related disorders, or a history of low blood pressure. In total, 20 participants were recruited, but 3 of them had low blood pressure, or even fainted, once standing up 5-10 minutes after insertion of the psoas electrode, leaving 17 participants. Their age was 28.7 ± 2.8 years (mean, SD), weight 60.7 ± 9.7 kg, height 167.6 ± 7.5 cm, and BMI 21.5 ± 2.4. Participants gave written informed consent. The protocol was approved by the local Medical Ethical Committee.

Electromyography (EMG) and kinematics

For fine-wire EMG, we used CE-marked paired hook-wire electrodes (40 gauge insulated stainless steel, VIASYS Healthcare, Madison WI, USA), threaded into sterile 50 mm or 100 mm hypodermic needles, with 5-7 mm-long "hooks" extending from the tip. For the transversus abdominis, these hooks were shortened manually to 2-3 mm. After disinfection, the needles were inserted with ultrasound guidance, under semi-sterile conditions. Insertion for the transversus abdominis was 2 cm medial to the midpoint of the vertical from the spina iliaca anterior superior (SIAS) to the rib cage (Hodges and Richardson, 1997); for the iliacus, 1 cm inferior to the inguinal ligament, 2 cm medial to the vertical from the SIAS; for the psoas major, starting 5-8 cm lateral to L3-L4 (Andersson et al., 1995). These procedures are regarded as safe (Katsavrias et al., 2005), but most of our subjects indicated that they were anxious about the psoas insertion (cf. Dörge et al., 1999), and in about 25%, the needle hit a transverse process, causing transient pain. During fast walking several subjects complained of discomfort from the iliacus electrodes. Any symptoms were transient and recovered after removal of the electrodes; no serious adverse effects were observed.

For surface EMG, pairs of electrodes (10 diameter Ag/AgCl discs, inter-electrode distance 20 mm; Kendall ARBO, Neustadt am Dom, Germany) were placed over the obliquus internus and externus abdominis, rectus abdominis, erector spinae, rectus femoris, adductor longus, biceps femoris, and gluteus maximus, following SENIAM recommendations (Hermens et al., 1999).

EMG data were amplified 20 times, band-pass filtered between 20 Hz and 1 kHz, and sampled at 2 kHz using a multichannel system (Porti, TMS-internationalTM, Enschede, The Netherlands) with input impedance adapted to the fine-wire. Only right-sided muscles were recorded.

Kinematic data were recorded with a 2 $\times$ 3 camera system (OPTOTRAK 3020, Northern DigitalTM, Waterloo, Ontario, Canada), connected via a synchronization cable to the Porti System. For ASLR, four cluster markers were attached to the upper and lower legs. For treadmill walking, markers were attached bilaterally to the calcaneus.

Conditions

In ASLR, subjects were supine, their legs straight and feet in dorsiflexion, 20 cm apart (Mens et al., 1999, 2001). They were instructed to raise each leg three times 20 cm with normal speed, without bending the knees, keeping the leg up for 10 s. The whole procedure was repeated with a non-elastic pelvic belt (3221/3300, Rafys, Hengelo, The Netherlands) around the pelvis, just below the SIAS ("high position", Damen et al., 2002; Mens et al., 2006b), with a tension of 50 N (Vleeming et al., 1992; Mens et al., 1999), fine-tuned with an inbuilt pressure gauge.

During walking, speed was increased from 1.4 km/h to 5.4 km/h (increments of 0.8 km/h). Each speed lasted two minutes, the second of which was recorded. The whole set of measurements was repeated with the pelvic belt, as described above.

Data analysis

Data were analysed with custom-made MATLAB 7.4 programs. Kinematic data were filtered with a 4th order bi-directional low pass Butterworth filter with a cut-off frequency of 5 Hz. For ASLR, onset and peak of leg raise were detected (first point with zero velocity before/after a peak in velocity), and leg raise velocity was derived (height of peak position divided by time to reach peak position from movement onset). From the markers in the relevant clusters, average heights of upper and lower legs were calculated over the three repetitions per subject per condition.

There were some missing values in fine-wire EMG (Tables 1 & 2). All EMG data were first high-pass filtered at 250 Hz (1st order Butterworth), to remove ECG contamination and to obtain more precise (less variable) estimates of the EMG amplitude (Potvin and Brown, 2004; Staudenmann et al., 2007), then full-wave rectified, and low-pass filtered at 5 Hz (2nd order Butterworth). Median EMG amplitude of the plateau during ASLR (5 to 10 s after movement onset) was calculated, then averaged over repetitions. For walking, we used median EMG activity per muscle per trial. Heel contacts were collected, and stride length calculated (as average stride time × treadmill speed).

Statistical analysis

Statistical analysis was performed with SPSS 16, using $p < 0.05$ as threshold for significance. To estimate the effects of Side and Condition in ASLR, and of Speed and Condition in walking (both including interaction), we used Generalized Estimation Equations (GEE), i.e., regression equations on the basis of repeated measures.

Table 1 Model *p*-values and regression coefficients (B) from GEEs of right-sided muscle activity (μV) during right or left ASLR ("Side"), with or without pelvic belt ("Condition"). Borderline significant values are given between brackets, and non-significant values have been left out

muscle	*N*	intercept		Side		Condition		interaction	
		p	*B*	p^a	B^b	p^a	B^c	p^a	B^d
Fine-wire EMG									
Transversus abdomnis	14	0.00	4.20	(0.09	1.32)	0.02	-1.13		
Iliacus	13	0.00	1.33	0.00	41.38	0.03	0.22	0.01	-5.31
Psoas	14	0.00	5.68						
Surface EMG									
Obliquus abdominis internus	17	0.00	1.44	0.00	0.47	0.00	-0.43		
Obliquus abdominis externus	17	0.00	1.09	0.02	0.16	(0.06	-0.24)		
Rectus abdominis	17	0.00	0.96						
Erector spinae	17	0.00	0.65						
Gluteus maximus	17	0.00	0.55	0.00	0.13				
Rectus femoris	17	0.00	0.77	0.00	7.01				
Adductor longus	17	0.00	0.88	0.00	2.01				
Biceps femoris	17	0.00	5.29	0.00	-4.40	0.00	0.66	0.00	0.71

Note that GEE calculates regression equations, and data in this Table should be read as, e.g., activity of the right m. iliacus (μV) equalled 1.33, plus 41.38 (for right-sided ASLR), plus 0.22 (for ASLR with a pelvic belt), minus 5.31 (for right-sided ASLR with a pelvic belt).

[a] These are *p*-values in the general model, not necessarily the same as *p*-values of the individual parametrizations

b For right-sided ASLR

c For ASLR with a pelvic belt

d For right-sided ASLR with a pelvic belt.

Results

ASLR

Heights reached by the upper/lower leg were not significantly affected by Side (right/left ASLR) or Condition (without/with pelvic belt), nor was there a significant Side $\times$ Condition interaction. The maximum velocity of leg raise was affected by Condition ($p < 0.001$), being faster with the belt (0.25 m/s vs. 0.23 m/s), while there was no significant effect of Side, or Side $\times$ Condition interaction.

All muscles showed task-dependent modulation of their activity (Fig. 1). Of the three muscles assessed with fine-wire (Fig. 2, upper panel), activity, in terms of voltage, was largest in the ipsilateral iliacus. With surface EMG (Fig. 2, lower panels), most activity was found in the ipsilateral rectus femoris, followed by the adductor longus. Contralaterally, these muscles were virtually inactive (for Side, p-values < 0.001). In contrast, activity of the biceps femoris was greater on the contralateral side (for Side, $p < 0.001$). Activity of the other muscles was essentially bilateral, but generally, ipsilateral activity was larger than contralateral activity (p-values < 0.03, but for the transversus abdominis this was borderline only, $p = 0.09$), while no effect of Side was found on the psoas major, rectus abdominis, and erector spinae (p-values > 0.4).

In ASLR (Table 1), the pelvic belt reduced activity in the transverse and oblique muscles of the abdominal wall, i.e., the transversus ($p = 0.02$), obliquus internus ($p < 0.001$), and obliquus externus abdominis (borderline, $p = 0.06$). In the specific parameterization we used, the iliacus ($p = 0.03$) and the biceps femoris ($p < 0.001$) were more active with the belt. Side $\times$ Condition interactions were significant for the iliacus ($p = 0.01$) and the biceps femoris ($p < 0.001$): The iliacus was most active ipsilaterally, but less so with the belt, while biceps femoris activity had the opposite pattern. Note that the negative effect (less activity) for right-sided iliacus activity with the belt is much larger than the positive general effect of the belt, so that, in fact, the most conspicuous effect of the pelvic belt on the m. iliacus is a reduction of activity during ipsilateral ASLR (Fig. 2).

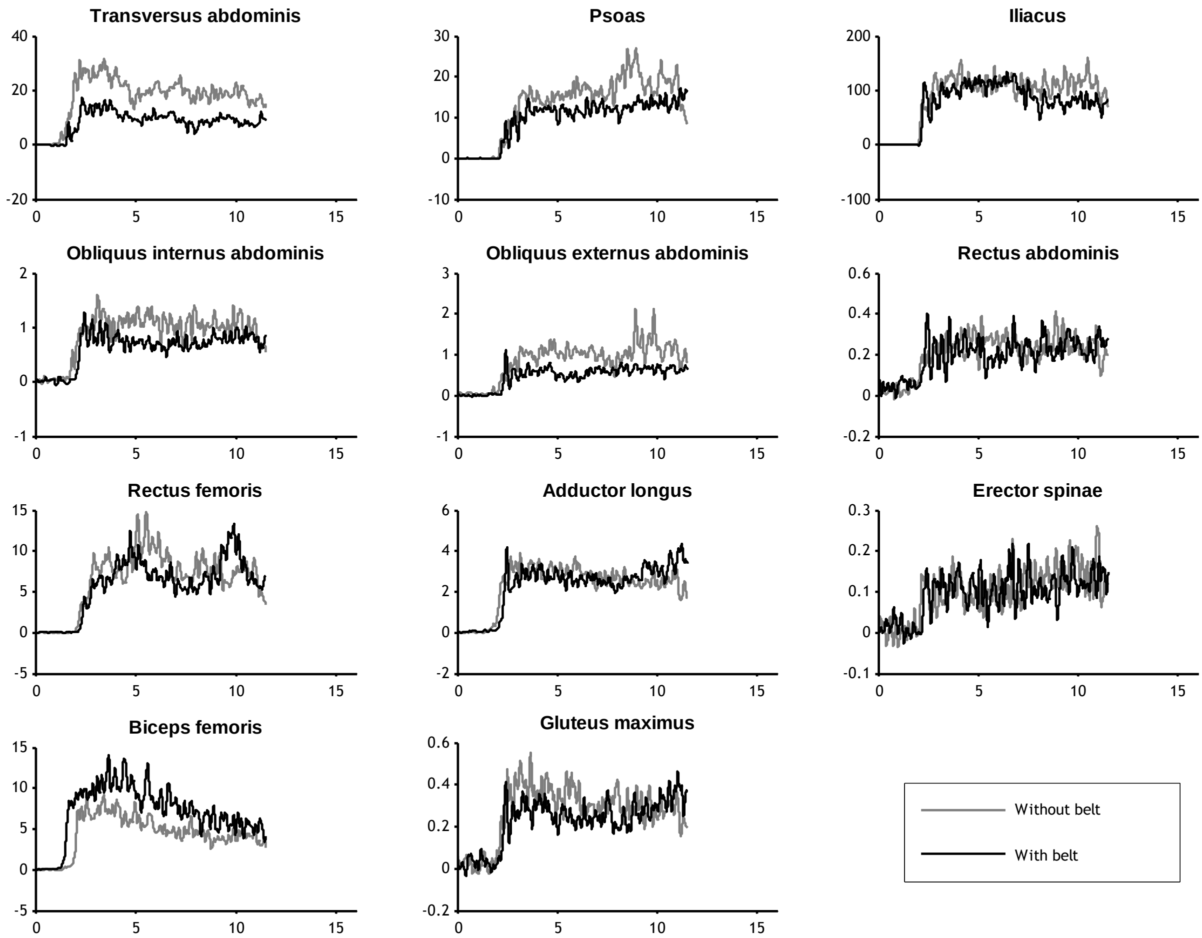

Fig. 1 Muscle activity during ASLR of one representative subject, each graph of one trial with (black), and one trial without (grey) pelvic belt. All activity is ipsilateral, except for the m. biceps femoris, where contralateral activity is depicted.

Treadmill walking

In treadmill walking, stride length increased with Speed (p < 0.001), but was not affected by Condition or by the Speed × Condition interaction. Most muscles (Table 2; Fig. 3) were more active at higher Speed (p-values < 0.001), but no effects of Speed were found for the iliacus, psoas, or biceps femoris, whereas the erector spinae appeared less active in faster walking (p = 0.06, borderline).

Table 2 Model p-values and regression coefficients (B) from GEEs of right-sided muscle activity (μV) during treadmill walking at six different velocities (from 1.4 km/h through 5.4 km/h, with increments of 0.8 km/h), with or without a pelvic belt ("Condition"). Borderline significant values are given between brackets, and non-significant values have been left out

muscle	*N*	intercept		Speed		Condition		interaction	
		p	*B*	*p*	B^a	*p*	B^b	*p*	B^c
Fine-wire EMG									
Transversus abdomnis	15	0.00	3.99	0.00	0.46	0.00	-1.08		
Iliacus	11	0.00	2.79			0.02	-0.92		
Psoas	16	0.00	3.10						
Surface EMG									
Obliquus abdominis internus	17	0.00	1.48	0.00	0.12	0.00	-0.51		
Obliquus abdominis externus	17	0.00	0.60	0.00	0.05	0.00	-0.07		
Rectus abdominis	17	0.00	0.52	0.00	0.04				
Erector spinae	17	0.00	1.20	(0.06	-0.03)	0.04	-0.10		
Gluteus maximus	17	0.00	0.57	0.00	0.03	0.04	0.16		
Rectus femoris	17	0.00	0.57	0.00	0.07			0.04	-0.01
Adductor longus	17	0.00	0.34	0.00	0.24				
Biceps femoris	17	0.00	0.81			0.01	-0.12		

For explanation, see Table 1.

[a] In the regression equation: *B* times the actual speed

[b] For treadmill walking with a pelvic belt

[c] For treadmill walking with a pelvic belt: *B* times the actual speed.

Discussion

ASLR involved activity of the abdominal wall, while both in ASLR and treadmill walking the pelvic belt led to less activity of the transverse and oblique abdominal muscles. Moreover, the contralateral m. biceps femoris was more active in ASLR with a belt, as was the m. gluteus maximus in treadmill walking with a belt. In ipsilateral ASLR, also activity of the m. iliacus was reduced.

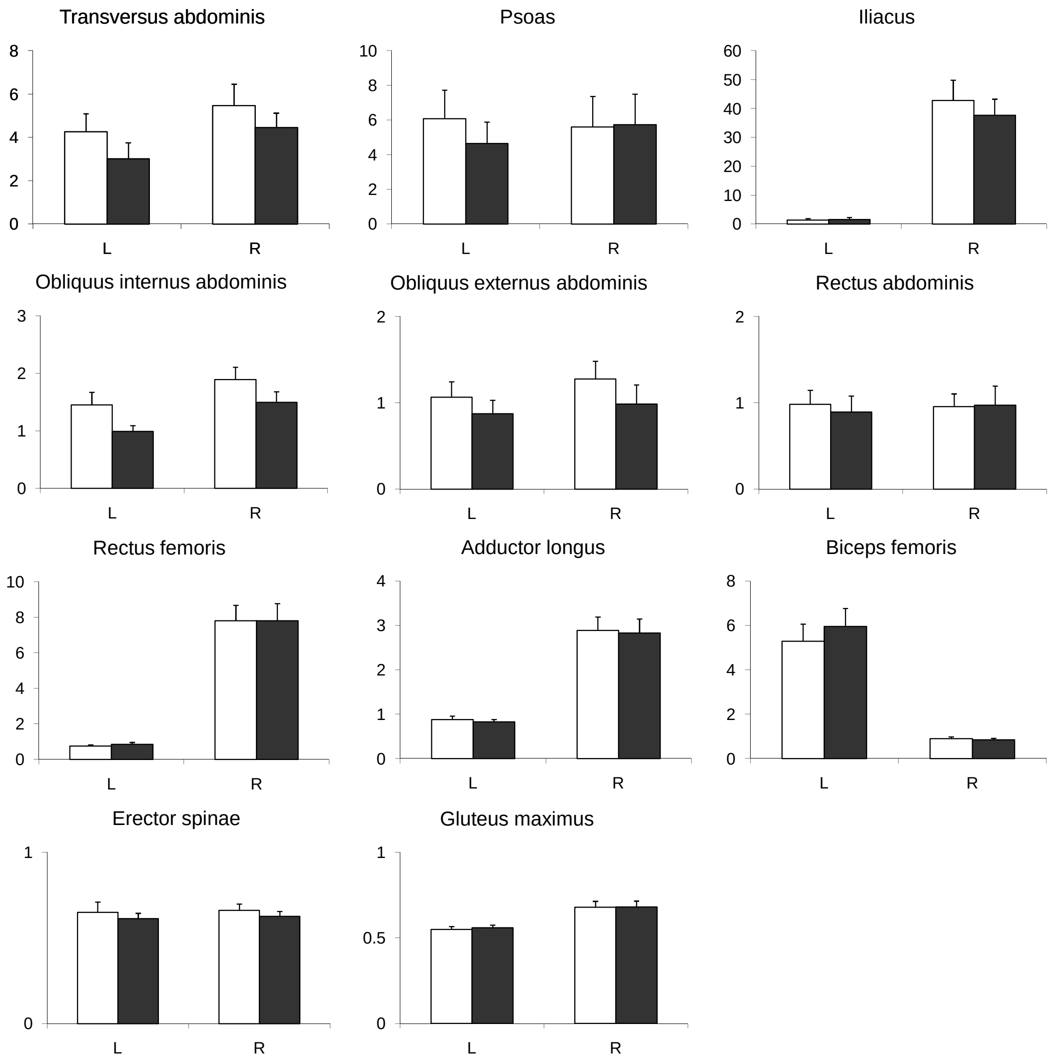

Fig. 2 Median activity, first averaged over three trials, then over all subjects, of right-sided muscles during left (L) and right (R) ASLR, without (white) and with (black) pelvic belt. Error bars represent standard errors. Note the scale differences.

Muscle activity in ASLR

ASLR is a complex movement. Prime movers appear to be the mm. iliacus, rectus femoris, and, maybe, adductor longus. In addition to raising the leg, the mechanical effect of these muscles is to pull the upper part of the ilium forward. In a study of ASLR in PPP patients, Mens et al. (1999) inferred from radiographs that the ilium actually rotated forward in these patients, which suggests that it was insufficiently stabilized. In our healthy subjects, abdominal wall muscle activity may have stabilized the ilium, thus offering the hip flexors a fixed origin. Abdominal wall activity was bilateral, but, with the exception of the m. rectus abdominis, more ipsilateral than contralateral (as also suggested in de Groot et al., 2008, Fig. 2; cf. Beales et al., 2009a). The preponderance of ipsilateral over contralateral activity appears to challenge (cf. Beales et al., 2009a) the old idea of balanced co-contraction in the oblique "muscle sling" of the m. obliquus abdominis internus on one side and the m. obliquus externus on the other (Benninghoff and Goerttler, 1964).

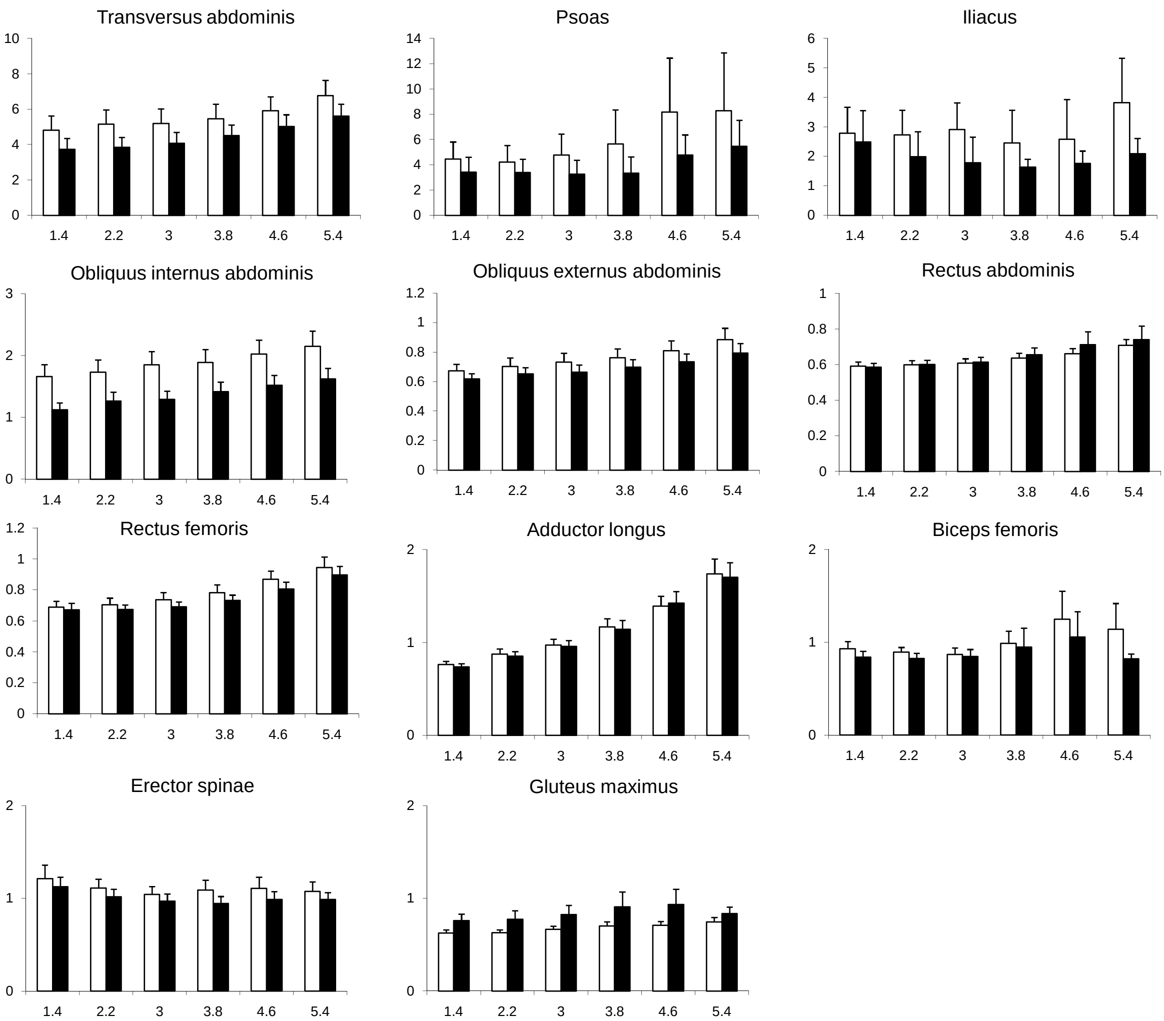

Fig. 3 Muscle activity (median values of the 1 minute recorded, averaged over all subjects) during treadmill walking at six different speeds, with (black) and without (white) pelvic belt. Error bars represent standard errrors.

When lifting one leg, subjects exerted contralateral downward pressure (Beales et al., 2009a) with their m. biceps femoris. The mm. rectus abdominis, psoas major, and erector spinae were more or less symmetrically active, which suggests that they were involved in the stabilization of the lumbar spine.

The transverse and oblique abdominal muscles behaved as predicted by Snijders' theory (Vleeming et al., 1990a & b). If these muscles were indeed delivering "force closure", with the pelvis moving as one unit in the sagittal plane, contralateral muscle activity (the abdominal wall and the m. biceps femoris) could be transferred easily to the ipsilateral iliac bone, thus preventing it to rotate forwards (Vleeming et al., 1992). This effect would have the additional benefit that m. iliacus fibres do not become too short (i.e., if these fibres were to shift to the far left of their length-tension curve, they would produce less force). Note that ipsilateral hip extensors would act against the iliacus around the hip joint, stiffening the joint, while these muscles cannot be used to stabilize the ipsilateral ilium relative to the sacrum.

Force closure and the pelvic belt

As predicted, the transverse and oblique abdominal muscles were less active during ASLR with a pelvic belt. Results for treadmill walking were similar. In ASLR, these effects were smaller for the m. obliquus abdominis externus, possibly because the fibbers of that muscle are more vertically oriented. Less activity of the m. obliquus abdominis internus was reported earlier for subjects standing upright with a pelvic belt (Snijders et al., 1998), but in that study, the m. transversus was not investigated. Hence, our results provide a powerful confirmation of Snijders' theory of "force closure" (Vleeming et al., 1990a & b; Snijders et al., 1993a & b), implying that coordinated muscular activity is needed to press the ilia against the sacrum. The pelvic belt appears to substitute for this stabilizing activity (Richardson et al., 2002; Pel et al., 2008).

The pelvic belt often relieves problems with ASLR (Mens et al., 1999 & 2006a), which suggests that patients who have difficulties performing ASLR may have problems with the active production of force closure. Why this would be so is beyond the scope of the present paper. Note that not all patients with positive ASLR have exactly the same problems (Mens et al., 1999), and not all patients with Pregnancy-related Pelvic girdle Pain (PPP), or groin pain, have trouble with ASLR (Mens et al., 2001, 2006a). Such heterogeneity requires further investigation, if only because it may explain, at least in principle, why no significant effect of a pelvic belt was found in RCTs on treatment forms for PPP that included a belt (Östgaard et al., 1994; Nilsson-Wikmar et al., 2005; Haugland et al., 2006; but see Vleeming et al., 2008).

Other effects of the pelvic belt

The actual effects of the pelvic belt went beyond the above. In ASLR, the task was performed faster with the belt. Subjects used more contralateral biceps femoris activity in ASLR, and more gluteus maximus activity in treadmill walking, both coinciding with less activity of the ipsilateral m. iliacus. In spite of lower iliacus activity, faster hip flexion probably coincided with a larger forward torque on the pelvis exerted by hip flexor muscles. This larger torque would need to be counteracted by a larger backward torque on the pelvis provided by hip extensors. In the ASLR, this can be done by contralateral biceps femoris activity, because its flexion moment around the knee is counteracted by the bench, but in walking this would need to be achieved by mono-articular hip extensor muscles. Note that it is unclear how the nervous system senses the presence of a pelvic belt, let alone how it selects strategies to perform hip flexion with the belt present. More research will be needed, with a focus on the actual kinematics of the iliac bone, and the role of the nervous system in selecting strategies.

Limitations

We studied young, healthy volunteers, who were aware of the purpose of our study, and generalization towards other populations will have to be established independently. All in all, we measured 11 muscles, which is more than in comparable studies, but still, far from complete. More important, we did not measure activity of the pelvic floor muscles (cf. Pool-Goudzwaard et al., 2005; Beales et al., 2009a). Since EMG voltages cannot be meaningfully compared between muscles or subjects, we could only use statistical analysis for within-subjects effects per muscle. Moreover, we initially attempted to register kinematics of the iliac bone, but then decided that removing the markers while placing the belt, and then putting the markers back, may have led to bias. Thus, we did not analyse pelvic movements.

Conclusion

Hip flexor muscles are prime movers in the Active Straight Leg Raise (ASLR). Activity of these muscles may cause forward rotation of the iliac bone, which is prevented by action of the abdominal wall muscles, and activity of contralateral hip extensors. Sacroiliac force closure by transverse and oblique abdominal muscles may facilitate transmission of contralateral torque to the ipsilateral iliac bone. A pelvic belt can also provide such force closure and leads to a reduction of the activity of transverse and oblique abdominal muscles. These insights may help to develop more precise diagnostics of patients with ASLR problems, or to better predict which patients may and which may not benefit from a pelvic belt. Still, we found several effects of the pelvic belt that could not be easily explained from within the framework of the present study.

Chapter 3

Is the psoas a hip flexor in the active straight leg raise?

Hu H, Meijer OG, Van Dieen JH, Hodges PW, Bruijn SM, Strijers RL, Nanayakkara PW, Van Royen BJ, Wu W, Xia C. European Spine Journal, 2011;20(5):759-65.

Abstract

Psoas function is a topic of considerable relevance in sports and clinical science. However, the literature on psoas function is not sufficiently consistent. Questions are, amongst others, if during hip flexion the psoas always has the same function as the iliacus, and if the psoas affects the hip more than the lumbar spine. In the present study, 17 healthy women, 20-40 years, performed the active straight leg raise (ASLR), with the right or the left leg ("Side"), and without or with weight added above the ankle ("Condition"). Electromyographic (EMG) activity of psoas and iliacus were recorded with fine-wire electrodes, and of rectus femoris and adductor longus with surface electrodes, all on the right side. Movements of the leg were recorded with active markers and a camera system. During ASLR, the iliacus, rectus femoris, adductor longus and psoas were active ipsilaterally, but psoas was also active contralaterally. All muscles started to contract before movement onset, the iliacus, rectus femoris, and adductor longus largely at the same time, before the psoas. There was no significant difference between the amplitude or time of onset of ipsilateral and contralateral psoas EMG activity, nor was there a significant interaction between Side and Condition for the psoas. Although ipsilateral psoas activity is consistent with the psoas being a hip flexor, contralateral activity is not. The most simplest explanation of the pattern found is that the psoas is bilaterally recruited to stabilize the lumbar spine, probably in the frontal plane.

Introduction

According to the 40th edition of Gray's Anatomy, "Psoas major …, together with iliacus, flexes the thigh …" (Standring, 2008, p. 1368). The text continues that psoas may be a lateral rotator of the hip, that bilateral psoas action bends the trunk and pelvis forwards, and that there is no evidence that unilateral psoas action causes lateral or forward flexion of the trunk, although in "symmetrical upright stance, iliopsoas has some action from below to maintain the vertebral column upright." (ibid). In sum, it is suggested that psoas mainly works with iliacus as a hip flexor, with also "some", that is to say: "less", effect on the spine. Theoretically and experimentally, however, these views are not without problems.

Psoas is a long muscle, with fascicles arising at T12–L5 from the bodies of two adjoining vertebrae and their intervertebral discs, as well as fascicles from the L1–L5 transverse processes (Bogduk et al., 1992). Psoas passes the pelvis, and inserts onto the trochanter minor. Owing to its position, it is difficult to investigate psoas non-invasively, and one early study of its function was performed in the framework of lumbar sympathectomy (Keagy et al., 1996). Since the 1990s, fine-wire electrodes have been inserted via a needle from the dorsal trunk under ultrasound guidance (e.g. Andersson et al., 1995), and the muscle became more accessible. Still, the literature on psoas function is rife with controversy and contradiction.

About half a century ago, Basmajian argued that psoas and iliacus could not be expected to have a different function, and inferred psoas activity from iliacus electromyography (EMG) (Basmajian, 1958). Since then, it has been shown that the two muscles "have individual and task specific activation patterns" (Andersson et al., 1995, p. 10). Still, even today, authors infer psoas function from surface EMG in the groin (e.g. De Groot et al., 2008). Another point of contradiction is Bogduk's view, a main source for Gray's Anatomy (Bogduk, 1997), that psoas cannot have much effect on the spine, because moment arms are small, and all fascicles have the same length. Bogduk (1992) concluded that psoas is "designed to act on the hip". However, this was disputed by Santaguida and McGill (Santaguida and McGill, 2005, p. 345), who argued that the anatomy of the psoas is "ideally suited" for lateral stabilization of the lumbar spine. Finally, most authors agree that psoas activity increases with larger hip flexion (Andersson et al., 1995), while Yoshio et al. (2002) even concluded that psoas mainly works as a stabilizer of the lumbar spine and the femoral head in the first 15° of hip flexion, and does not become an effective hip flexor before 45° of flexion.

The above debates are far from trivial. Psoas function has drawn much attention in sports (Hides et al., 2010) and clinical literature (Juker et al., 1998; Marrè-Brunenghi et al., 2008). The present study focuses on psoas function during the active straight leg raise (ASLR). The ASLR involves hip flexion, but also challenges the stability of the lumbar spine due to the large moment of gravity and the effects of muscles (Hu et al., 2010). Investigation of the ASLR is also clinically relevant, as ASLR is often limited in pregnancy-related pelvic girdle pain (Wu et al., 2004a).

Methods

Participants were healthy 20–40 years old women with normal blood pressure. In total, 20 participants were recruited. Three participants became light-headed or fainted once standing up 5–10 min after insertion of the psoas electrode, leaving 17 participants. The age, weight, height and BMI of the participants were 28.7 ± 2.8 years (mean, SD), 60.7 ± 9.7 kg, 167.6 ± 7.5 cm, and 21.5 ± 2.4, respectively. Participants gave written informed consent. The protocol was approved by the Local Medical Ethical Committee.

Electromyography

Muscles were measured on the right side only. For fine-wire EMG, we used CE-marked paired hook-wire electrodes (25 gauge insulated stainless steel, VIASYS Healthcare, Madison WI, USA), threaded into sterile 50 or 100 mm hypodermic needles, with 5–7 mm long “hooks” extending from the tip. After disinfection, the needles were inserted with ultrasound guidance, under semi-sterile conditions. Iliacus insertion was 1 cm inferior to the inguinal ligament, 2 cm medial to a vertical line down from the anterior superior iliac spine, and for the psoas major, 5–8 cm lateral to L3–L4 (Andersson et al., 1995). The use of ultrasound allowed visual monitoring of the final placement of the fine-wire electrodes. These procedures are regarded as safe, but most of our subjects were anxious about psoas insertion and in about 25%, the needle hit a transverse process, causing some pain (Hu et al., 2010). Any symptoms were transient and recovered after the removal of the electrodes; no serious adverse effects were observed.

For surface EMG, pairs of electrodes (10 diameter Ag/AgCl discs, inter-electrode distance 20 mm; Kendall ARBO, Neustadt am Dom, Germany) were placed over the rectus femoris and adductor longus (Hu et al., 2010).

EMG data were amplified 20 times, band-pass filtered between 20 Hz and 1 kHz, and sampled at 2 kHz using a multichannel system (Porti, TMS-International, Enschede, The Netherlands), with input impedance adapted to the fine wire.

Kinematics

Four cluster markers were attached to the upper and lower legs. Each cluster marker contained three infrared emitting diodes for movement registration with a 2 × 3 camera system (OPTOTRAK 3020, Northern Digital, Waterloo, ON, Canada), connected via a synchronisation cable to the Porti. For kinematics, the sampling frequency was 50 Hz.

Conditions

In right and left ASLR, subjects were supine, their legs straight and feet in dorsiflexion, 20 cm apart (Mens et al., 1999). They were instructed to raise each leg three times until the foot reached 20 cm above the table, without bending the knees, and keeping the leg elevated for 10 s.

The whole procedure was repeated with weight added just above the ankle. Using lower extremity anthropometry and specific regression equations (Zatsiorsky, 2002, p. 605), the mass of this weight was calculated in such a way that the static moment of the leg with respect to the hip was increased by 50%.

Data analysis

Data were analysed with MATLAB 7.4 (The Mathworks, Nattick, MA, USA). Kinematic data were filtered with a 4th order bi-directional low-pass Butterworth filter with a cutoff frequency of 5 Hz. Onset and peak of leg raise were derived (first point with zero velocity before/after a peak in velocity), as was leg raise velocity (height of peak position divided by time to reach peak position). From the markers in the relevant clusters, average heights of the upper and lower legs were calculated over the three repetitions per subject per condition.

Fine-wire EMG was not usable from four subjects. Further, two sets of iliacus data contained too much noise, particularly during the condition with weight added; these data were removed from amplitude calculations (leaving, for amplitude, N = 13 for the psoas, iliacus N = 11, rectus femoris and adductor longus N = 17). The onset signal was too noisy in three sets of psoas data, and these were removed from onset calculations. Synchronisation failed to work three times, rendering onset determination impossible, which led to the removal of one more set of psoas data, and three sets of rectus femoris and adductor longus data (leaving, for onset, N = 9 for the psoas, iliacus N = 11, rectus femoris and adductor longus N = 14).

All EMG data were high-pass filtered bi-directionally at 250 Hz (first-order Butterworth), to remove ECG contamination and to obtain more precise (less variable) estimates of the EMG amplitude (Potvin and Brown, 2004; Staudenmann et al., 2007), then full-wave rectified, and low-pass filtered bi-directionally at 5 Hz (second-order Butterworth). Median EMG amplitude during plateau of movement (5–10 s after movement onset) was calculated. Median onset time was determined by means of a dedicated algorithm (using the approximated generalised likelihood ratio (Staude and Wolf, 1999), where necessary corrected on the basis of visual inspection (Hodges and Bui, 1996). EMG onset was expressed in relation to movement onset.

Statistical analysis

Statistical analysis was performed with SPSS 16, with $p < 0.05$ as threshold for significance. Generalised estimation equations (GEEs) on the amplitude of muscle activity were used, with Side (ipsilateral vs. contralateral) and Condition (without vs. with weight added) as factors. Then, GEEs were performed on onset time of muscle activity, for all relevant muscle pairs, with Muscle (the one vs. the other in the pair in question) and Condition (without vs. with weight added) as factors. GEEs calculate regression equations on the basis of repeated measures, while allowing for missing values.

Results

Heights reached by the upper/lower leg were not significantly affected by Side (right/left ASLR) or Condition (without/with weight), nor was there a significant Side × Condition interaction. Further, the velocity of the leg raise was not significantly affected by Side, Condition, or their interaction.

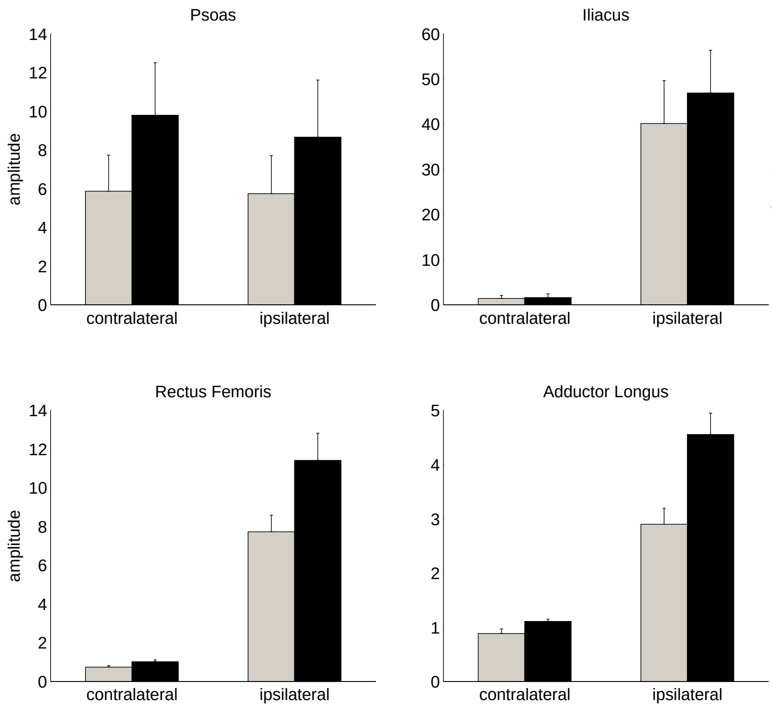

Fig. 1 EMG activity (μV) of the psoas, iliacus, rectus femoris, and adductor longus, during contralateral and ipsilateral ASLR without (grey) or with (black) weight added. Error bars represent standard deviations

Amplitude of muscle activity

During right or left ASLR (Fig. 1), there was ipsilateral activity in all four muscles. Psoas was also active during contralateral ASLR.

In GEEs on amplitude (Table 1), an effect of Side was found for iliacus, rectus femoris, and adductor longus (p = 0.00), with more activity during ipsilateral ASLR. There was no effect of Side for the psoas (p = 0.98). All four muscles were significantly more active with weight (p = 0.00). For iliacus, rectus femoris, and adductor longus, there were significant Side × Condition interactions (p = 0.00), with more activity during ipsilateral ASLR with weight added. No such interaction was found for psoas (p = 0.95).

Table 1 Regression coefficients (B) and *p* values from GEEs on amplitudes of right-sided muscle activity during the active straight leg raise (ASLR), with Side (ipsilateral vs. contralateral) and Condition (without vs. with weight added above the ankle) as factors

amplitude (εV)	intercept		Side		Condition		interaction	
	B	p^a	B^b	p^a	B^c	p^a	B^d	p^a
psoas	6.41	0.00	-0.08	0.98	3.98	0.00		
iliacus	1.44	0.00	38.66	0.00	0.15	0.00	6.61	0.00
rectus femoris	0.74	0.00	6.99	0.00	0.28	0.00	3.41	0.00
adductor longus	0.88	0.00	2.01	0.00	0.23	0.00	1.43	0.00

Non-significant interactions have been left out

Note that GEE calculates regression equations, and, for instance, the first line reads as: psoas activity equals 6.41 + (in conditions with weight) 3.98 (and some non-significant components).

a. for the model effects (which may be different from *p*-values for specific parameterizations)
b. for the ipsilateral side (that is, right side ASLR)
c. for the condition with weight added
d. ipsilateral, with weight

Table 2 Regression coefficients (B) and *p* values from GEEs on amplitudes of right-sided muscle activity during the active straight leg raise (ASLR), with Side (ipsilateral vs. contralateral) and Condition (without vs. with weight added above the ankle) as factors

	Intercept		Muscle		Condition		interaction	
comparison[a]	*B*	p^b	B^c	p^b	B^d	p^b	B^e	p^b
Ipsilateral versus contralateral:								
PS vs. PS	-117.94	0.00	53.63	0.15	-61.74	0.00		
Ipsilateral versus ipsilateral:								
PS vs. IL	-118.52	0.00	-68.60	0.01	-47.09	0.00		
PS vs. RF	-116.99	0.00	-56.51	0.01	-51.19	0.00		
PS vs. AL	-106.10	0.00	-74.27	0.01	-67.28	0.00		
IL vs. RF	-195.88	0.00	14.46	0.47	-35.32	0.03		
IL vs. AL	-193.79	0.00	14.68	0.62	-31.59	0.04	-38.23	0.02
RF vs. AL	-172.07	0.00	-14.93	0.37	-54.04	0.00		

Cf. also Table 1

a. PS: m. psoas
IL: m. iliacus
RF: m. rectus femoris
AL: m. adductor longus
b. for the model effects (which may be different from *p*-values for specific parameterizations)
c. for the muscle mentioned last in the first column
d. for the condition with weight added
e. for the muscle mentioned last in the first column, in conditions with weight added

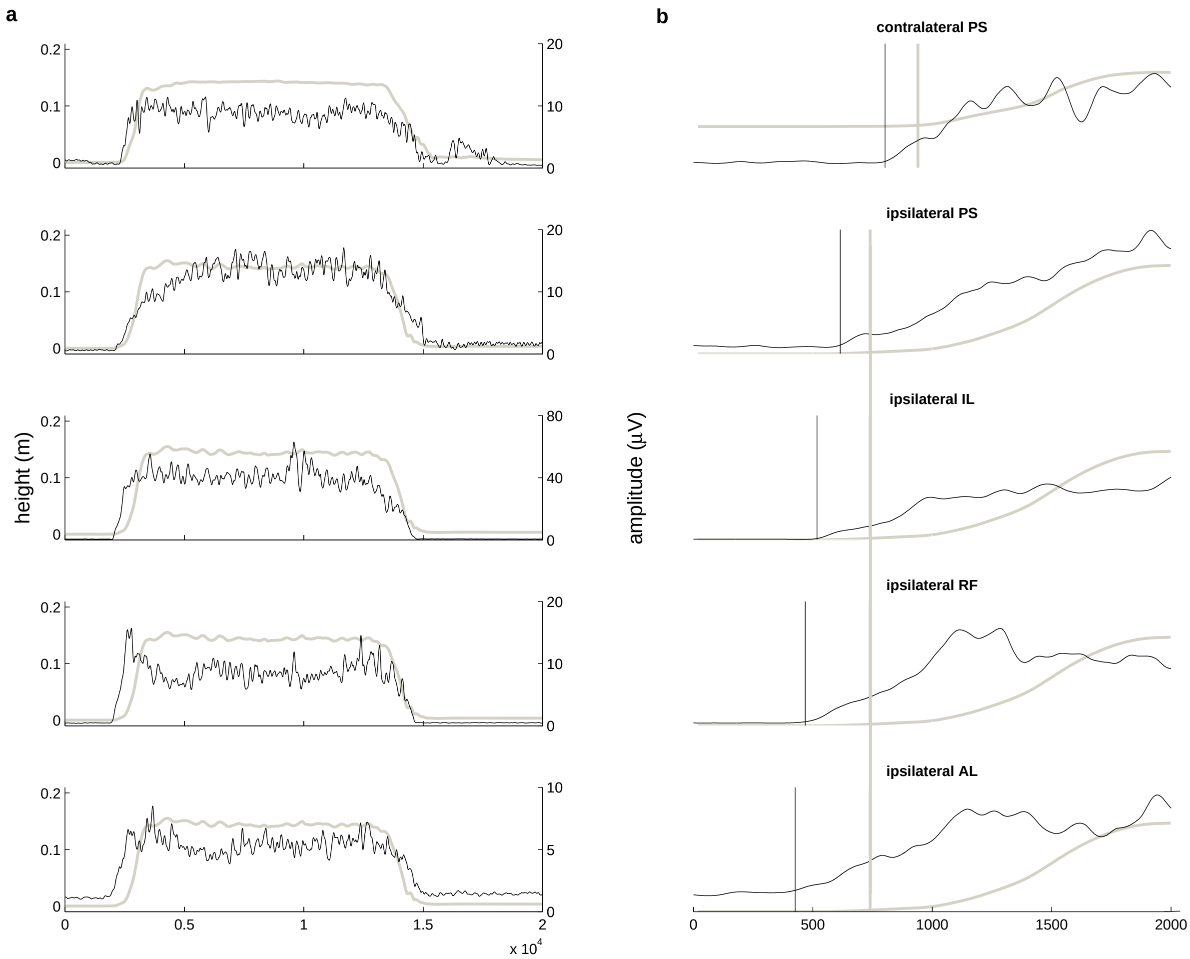

Fig. 2 a Representative example, in a condition with weight added, of muscle activity (µV, right vertical axis, drawn in black) during ASLR. Movement of the lower leg is also depicted (m, left vertical axis, drawn in grey). Time (ms) is given on the horizontal axis. For muscle names and side (see Fig. 2b). **b** Greater temporal resolution of the onset of muscle activity (black vertical lines), and the onset of movement of the lower leg (grey vertical lines). PS psoas, IL iliacus, RF rectus femoris, AL adductor longus

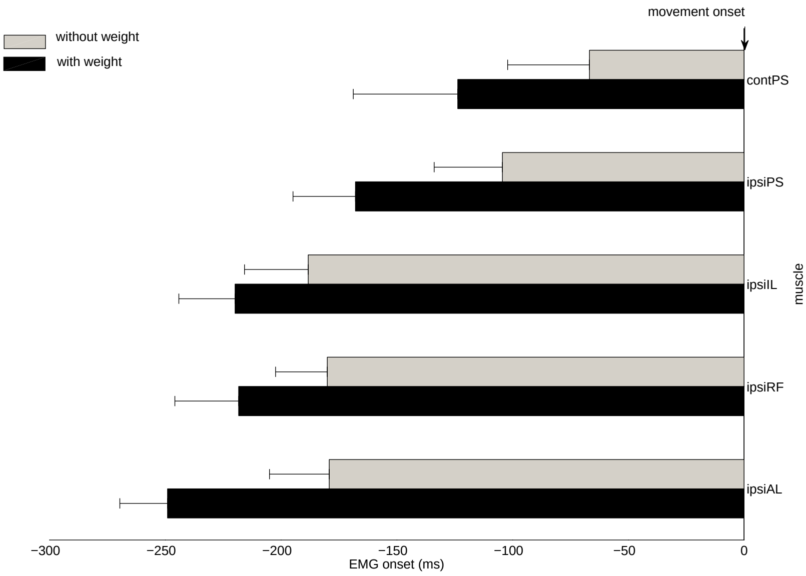

Fig. 3 Time of onset of EMG activity during the active straight leg raise (ASLR), before the onset of elevation of the lower leg, in the contralateral psoas (contPS), the ipsilateral psoas (ipsiPS), the ipsilateral iliacus (ipsiIL), the ipsilateral rectus femoris (ipsiRF), and the ipsilateral adductor longus (ipsiAL), in conditions without (grey) and with (black) weight added

EMG activity typically increased before onset of leg elevation (Figs. 2 & 3). GEEs of onset time (Table 2), ipsilaterally, revealed that iliacus, rectus femoris, and adductor longus EMG increased before that of psoas (muscle, $p = 0.01$). Onset of ipsilateral iliacus, rectus femoris, and adductor longus EMG was largely synchronous, with one exception, adductor longus EMG increased before iliacus in the condition with weight ($p = 0.02$). There was no significant difference between ipsilateral and contralateral psoas onset times ($p = 0.15$). EMG onset of all muscles was earlier when weight was added ($p \leq 0.04$).

Discussion

Muscle activity was recorded of psoas and iliacus (fine-wire EMG electrodes), as well as rectus femoris and adductor longus (surface EMG electrodes), during an ASLR, without or with a weight added above the ankle. All muscles showed task-related activity. Iliacus, rectus femoris, and adductor longus were active during ipsilateral ASLR only, whereas psoas was also contralaterally active. Onset of muscle activity preceded leg elevation, but was somewhat later for psoas than for the other three muscles. Muscles were more active, and started their activity earlier, when weight was added. Psoas EMG amplitude and onset time were, on average, not different between ipsilateral and contralateral ALSR, and there was also no interaction between Side (ipsilateral vs. contralateral) and Condition (without or with weight added).

Hip flexion

As psoas passes anterior to the flexion/extension axis of the hip on its way to the trochanter minor, mechanically, psoas is a hip flexor. In principle, this is consistent with the psoas activity found during ipsilateral ASLR. However, psoas was also active during contralateral ASLR, with similar amplitude and onset time. This pattern did not change when weight was added above the ankle. These results imply that psoas is not used as a hip flexor contralaterally. Rather than invoking different explanations for ipsilateral and contralateral psoas activity, the simplest account would be that psoas is recruited bilaterally for some other function than hip flexion (Santaguida and McGill, 1995).

Juker et al. (1998) studied psoas activity in a wide variety of tasks, and the muscle was found to be most active in standing, when raising the ipsilateral knee and flexing the hip, while the upper leg was pushed down with the hands. Unfortunately, however, contralateral activity was not measured. In the present study, the psoas was also active contralaterally.

Andersson et al. (1995) studied psoas activity in seven subjects, who performed a large number of tasks. The psoas was found to be active during an ipsilateral dynamic straight leg raise, up and down 60°, but silent contralaterally. This result appears to be in contrast with the present study. Psoas recruitment is task-dependent (Andersson et al., 1997), and maybe the dynamic straight leg raise up to 60° is a different task from the ASLR in the present study. Perhaps, there are inter-individual differences in the details of motor control, task performance, and/or initial position. Post hoc we inspected psoas activity signals of the 13 subjects with high quality EMG data. In three subjects, there was more ipsilateral than contralateral psoas activity, in four ipsilateral activity was less, and in six activities appeared equal. Still, such inter-individual variability cannot explain the difference with the study of Andersson et al. (1995), who reported no contralateral activity, while the present study found contralateral activity in all subjects.

Spinal stabilization

Because the psoas connects the lumbar spine with the femur, it can affect not only the hip joint, but also the lumbar spine, and, more indirectly, the pelvis. During ASLR, the psoas may be recruited to stabilize the lumbar spine.

Nachemson (1966,1968) suggested that psoas contraction stabilizes the spine, thereby adding to compressive forces, which stiffen spinal joints (Garnder-Morse and Stokes, 2003). In the frontal plane, psoas can be used for lateral stabilization (Andersson et al., 1995), and bilateral psoas activation appears to work as guy wires stabilizing the spine as if it were a mast of a ship (Santaguida and McGill, 1995). In the sagittal plane, psoas might contribute to the regulation of lumbar lordosis (Penning, 2000), and in the transverse plane, the role of psoas in spinal stabilization is dependent upon the exact parameters of the task at hand (Andersson et al., 2002).

If subjects raise a leg from the supine position, gravity will exert a considerable moment, and many muscles will be involved (Hu et al., 2010). Thus, ASLR may perturb the lumbar spine. Although the mechanisms to stabilize the lumbar spine are outside the scope of the present study, it appears plausible that bilateral psoas activation during ASLR served to stabilize the lumbar spine, particularly in the frontal plane. Nikolai Bernstein once wrote: “There are no situations in which muscle shortening is the cause of a movement” (Fel’dman and Meijer, 1999, p. 119), highlighting that one always has to take the whole interplay of forces into account. In ASLR, psoas activity cannot be understood unless, amongst other factors, gravity is accounted for, the activity of other muscles, and the need to stabilize the lumbar spine.

Limitations

The determination of EMG onset time is sensitive to noise, and in the present study, other technical problems were also encountered, probably related to instability of connections. Nevertheless, all usable psoas signals showed contralateral activity, and the results of statistical analysis clearly suggested that psoas activity was bilateral. It is highly unlikely that technical problems systematically affected our main conclusions.

Fine-wire EMG records activity from a small region of the muscle, in the present study at the level of L3–L4 (Andersson et al., 1995), and it may be desirable to replicate the study for other parts of the psoas. More important, the range of hip flexion was limited in ASLR, well below the 45° where hip flexion was argued to become a major function of psoas (Yashio et al., 2002). It will be important to replicate the present study with larger amounts of hip flexion.

Initial trunk position was not controlled, and possible frontal plane deviations in the kinematics of the raising leg were not recorded. These factors may have contributed to inter-individual variation.

Conclusion

In the ASLR, the ipsilateral iliacus, rectus femoris, adductor longus, and psoas are active, as is the contralateral psoas. Contralaterally, the psoas cannot be used as a hip flexor. The psoas is probably active bilaterally to stabilize the lumbar spine in the frontal plane.

Chapter 4

Control of the lateral abdominal muscles during walking

Hu H, Meijer OG, Van Dieen JH, Hodges PW, Bruijn SM, Strijers RL, Nanayakkara PW, Van Royen BJ, Wu W, Xia C. Human Movement Science, under revision.

Abstracts

Transversus abdominis (TA), obliquus internus (OI), and obliquus externus (OE) are involved in multiple functions: breathing, control of trunk orientation, and stabilization of the pelvis and spine. How these functions are coordinated has received limited attention. We studied electromyographic (EMG) activity of right-sided muscles and 3-dimensional moments during treadmill walking at six different speeds (1.4-5.4 km/h) in sixteen healthy young women. PCA revealed time series of trunk moments to be consistent across speeds and subjects though somewhat less in the sagittal plane. All three muscles were active during ≥ 75% of the stride cycle, indicative of a stabilizing function. Clear phasic modulations were observed, with TA more active during ipsilateral, and OE during contralateral swing, while OI activity was largely symmetrical. Fourier analysis revealed four main frequencies in muscle activity: respiration, stride frequency, step frequency, and a triphasic pattern. With increasing speed, the absolute power of all frequencies remained constant or increased; the relative power of respiration and stride-related activities decreased, while that of step-related activity and the triphasic pattern increased. Effects of speed were gradual, and EMG linear envelopes had considerable common variance (> 70%) across speeds within subjects, suggesting that the same functions were performed at all speeds. Maximum cross-correlations between moments and muscle activity were 0.2-0.6, and further analyses in the time domain revealed both simultaneous and consecutive task execution. To deal with conflicting constraints, the activity of the three muscles was clearly coordinated, with co-contraction of antagonists to offset unwanted mechanical side-effects of each individual muscle.

Introduction

The human body has three lateral abdominal muscles: transversus abdominis (TA), obliquus internus (OI), and obliquus externus (OE). Their anatomy suggests a range of functions. In the frontal plane, unilateral OI and OE activity may be involved in ipsilateral bending of the trunk (Standring, 2008). In the transverse plane, OE contributes to contralateral rotation, whereas OI and TA contribute to ipsilateral rotation (Benninghoff & Goerttler, 1964; Urquhart & Hodges, 2005; Hodges, 2008). In the sagittal plane, bilateral activity of the lateral abdominal muscles can contribute substantially to trunk flexion (McGill, 1996).

Bilateral activity of the lateral abdominal muscles may also be involved in stability. Bilateral TA activity was suggested to stabilize the lumbar spine (Hodges & Richardson, 1997a; Hodges, 1999), and concerted action of the lateral abdominal muscles can press the iliac bones against the sacrum, thus providing pelvic stability ("force closure", cf. Vleeming, Volkers, Snijders & Stoeckart, 1990; Snijders, Vleeming & Stockart, 1993; Hu et al., 2010).

Together with the rectus abdominis, the lateral abdominal muscles form the abdominal wall. Continuous activity of the abdominal wall can be seen as holding the viscera; it regulates intra-abdominal pressure, which implies a role in breathing (Standring, 2008). Finally, activity of the abdominal wall, plus the diaphragm and the pelvic floor, contributes to spinal stability (Hodges, Butler, McKenzie & Gandevia, 1997; Hodges, Eriksson, Shirley & Gandevia, 2005; Cholewicki, Juluru, Radebold, Panjabi & McGill, 1999; Pel, Spoor, Pool-Goudzwaard, Hoek van Dijke & Snijders, 2008).

Several studies have attempted to verify the roles of the lateral abdominal muscles through electromyographic recordings during the performance of specific tasks. In subjects holding loads, OE was involved in lateral bending (Seroussi & Pope, 1987). OI and OE played a role in trunk rotation during twisting efforts against resistance (McGill, 1991), and TA was active during ipsilateral rotation (Urquhart & Hodges, 2005). Considerable activity of all three muscles was reported in flexion exercises (Juker, McGill, Kropf & Steffen, 1998). OE was involved in expiration (Campbell, 1952), and the activity of all three muscles was modulated at the frequency of breathing (Saunders, Schache, Rath & Hodges, 2004). Bilateral TA activity preceded rapid movements of upper and lower extremities, which supports the idea that TA stabilizes the spine (Hodges & Richardson, 1996, 1997a, 1997b). Furthermore, the use of a pelvic belt led to decreased EMG amplitudes of these muscles during the Active Straight Leg Raise, in accordance with their assumed function in sacroiliac stabilization (Hu et al., 2010).

As a rule, the mechanical effects of any single muscle will not coincide with the exact 3-dimensional demands of a task, and when constraints are conflicting, there will be no perfect solution. More important, the lateral abdominal muscles may subserve multiple functions simultaneously (Benninghoff & Goerttler, 1964; Hodges, 2008). Some evidence for such multitasking has been provided for trunk postural control plus breathing (Hodges, Gurfinkel, Brumagne, Smith & Cordo, 2002; Saunders et al., 2004). Still, how multiple functions are coordinated has received limited attention only. In multitasking, or dealing with conflicting constraints, the control system may prioritize one function and ignore the other (cf. Hodges, Heijnen & Gandevia, 2001). Also, the control system may exploit muscle redundancy by assigning specific functions to specific muscles or parts thereof (Puckree, Cerny & Bishop, 1998). Finally, problems of multitasking may be alleviated when the system can deal with different task demands consecutively (task switching) rather than simultaneously (multitasking).

The present study focused on control of the lateral abdominal muscles during gait, which is an activity that involves all functions mentioned above (cf., e.g., Saunders, Inman & Eberhart, 1953; Callaghan, Patla & McGill, 1999; Saunders et al., 2004). We used electromyography (EMG), calculated trunk moments using a dynamic 3-dimensional linked segment model, and cross-correlated EMG linear envelopes with moment data. The stride cycle was split into different phases (cf. Perry, 1992; Ivanenko, Poppele & Lacquaniti, 2004), with peak detection in the time series of moments and of EMG. Fourier analysis of the EMG linear envelopes was performed to determine the absolute and relative power at main frequencies (Saunders et al., 2004). Since the relative importance of different functions changes with speed (Saunders et al., 2004; Anders et al., 2007), walking speed was manipulated.

The goal of this study was to understand the coordination of the lateral abdominal muscles during gait. We expected that these muscles would be engaged in multitasking most of the time. Because normal walking is relatively undemanding, we expected that no function would be ignored, and, given the literature (Hodges, 2008), that no (part of a) specific muscle would be specifically dedicated to a single task. On the contrary, we hypothesized that the control system co-activates muscles to produce the desired effect, dealing with conflicting constraints by co-contraction to offset unwanted mechanical side-effects of any individual muscle.

Methods

Subjects

Sixteen healthy, nulliparous females were enrolled (mean ± SD age 27.5 ± 2.7 years, weight 61.2 ± 9.8 kg, height 167.9 ± 7.6 cm, BMI 21.6 ± 2.4 kg/m2). Exclusion criteria were: previous orthopaedic surgery, walking-related disorders, or a history of low blood pressure. Participants gave written informed consent. The protocol was approved by the local Medical Ethical Committee.

Data collection

Kinematic and electromyographic data were collected during treadmill walking at six different speeds (1.4-5.4 km/h, with increments of 0.8 km/h). Each speed condition lasted two minutes, the second of which was recorded. Kinematic data were recorded at 50 samples/s with a 2×3 camera system (Optotrak 3020, Northern DigitalTM, Waterloo, Ontario, Canada), connected via a synchronization cable to an EMG data collection system (Porti, TMS-internationalTM, Enschede, The Netherlands; CMRR > 90 dB, 22 bits AD conversion after $20\times$ amplification, resolution 71.5 nV/bit), with input impedance adapted for fine-wire EMG. EMG data were band-pass filtered between 20 Hz and 1000 Hz, and sampled at 2000 samples/s.

For kinematic registration, clusters of three LED Markers were fixed onto small lightweight custom-made triangular frames, and attached to the heels of both shoes, the pelvis (between the posterior superior iliac spines), the thorax (Th6), upper arms, forearms and head. Before the experiment, the Optotrak system was calibrated, and cluster markers were related to anatomical landmarks (Cappozzo, Catani, Croce & Leardini, 1995), using a 6-marker pointer. Subjects acclimatized for several minutes on the treadmill at the six different speeds.

TA EMG activity was recorded with intramuscular fine-wire electrodes (CE-marked paired hook-wire, 25 gauge insulated stainless steel, VIASYS Healthcare, Madison WI, USA), threaded into sterile 50 mm hypodermic needles, and trimmed, with 2-3 mm long "hooks" extending from the tip. After disinfection, the needle was inserted under ultrasound guidance, in semi-sterile conditions. The insertion point was 2 cm medial to the midpoint of the vertical from the spina iliaca anterior superior to the rib cage (Hodges & Richardson, 1997a). Some subjects felt anxious and uncomfortable when the needle entered the muscle, but no lasting pain was reported.

For OI and OE, EMG was recorded with pairs of surface electrodes (24 mm diameter Ag/AgCl discs, inter-electrode distance 20 mm; Kendall ARBO, Neustadt am Dom, Germany), following SENIAM recommendations (Hermens et al., 1999).

All muscles were recorded on the right side of the body only.

Data analysis

All data were analysed with MATLAB 7.4 (The MathWorks, Natick, MA, USA). Kinematic data were filtered with a 4th order bi-directional low pass Butterworth filter with a cut-off frequency of 5 Hz. Heel contacts were derived from the minimum vertical position of the average of the three LEDs on the heel. The stride cycle was defined from one ipsilateral (right) heel contact to the next. Stride frequency (Hz) was calculated per speed as 1 divided by the average time (s) between consecutive ipsilateral heel contacts. Step length (m) was calculated as average step time (s) between alternate heel contacts, multiplied by treadmill speed (m/s). The between-subjects standard-deviations of stride frequency and step length were calculated per speed (Table 1).

To construct a phase map, stride cycles were divided into eight phases, starting with ipsilateral heel contact (cf. Perry, 1992): 1) contralateral (left) pre-swing (PSw, 0-10% of the stride cycle, largely consisting of double support); 2) contralateral initial swing (ISw, 10-25% of the stride cycle); 3) contralateral mid-swing (MSw, 25-35%); 4) contralateral terminal swing (TSw, 35-50%), and phases 5 through 8 repeating the corresponding sequence for ipsilateral swing.

Table 1 Average ± standard-deviation of stride frequency and step length.

Speed	1.4 km/h	2.2 km/h	3.0 km/h	3.8 km/h	4.6 km/h	5.4 km/h
Frequency (Hz)	0.52 ± 0.08	0.66 ± 0.07	0.76 ± 0.05	0.85 ± 0.04	0.93 ± 0.04	0.99 ± 0.05
Step length (m)	0.38 ± 0.03	0.47 ± 0.02	0.55 ± 0.02	0.62 ± 0.02	0.69 ± 0.02	0.76 ± 0.02

To estimate moments at the L5S1 joint, a dynamic 3-dimensional linked segment model was used (Kingma, Toussaint, De Looze & Van Dieën, 1996), with the moment around X representing trunk lateral bending, Y flexion and extension, and Z rotation. The kinematics of the upper body (pelvis, thorax, hand with forearms, upper arms, and head), the estimated masses, and the inertia tensors of the upper body segments were used as inputs in a global equation of motion (Hof, 1992), and projected onto a pelvis axis system. Time series of moments were time-normalized per stride (0-100%), and averaged over strides per time point per subject per speed. These curves were then averaged over subjects, and graphically depicted (Fig. 1). The within-stride standard deviation of the average stride cycle per subject per speed was used for statistical analysis of the modulation of the moment over the stride cycle.

Table 2 Average ± standard-deviation of percentage of time active[1] of the transversus abdominis (TA), obliquus internus (OI) and obliquus externus (OE).

Speed	1.4 km/h	2.2 km/h	3.0 km/h	3.8 km/h	4.6 km/h	5.4 km/h
TA (%)	90.6 ± 0.16	92.3 ± 0.07	91.3 ± 0.11	92.6 ± 0.09	93.4 ± 0.09	95.9 ± 0.06
OI (%)	96.8 ± 0.09	97.1 ± 0.08	96.6 ± 0.10	97.7 ± 0.08	97.8 ± 0.08	97.5 ± 0.08
OE (%)	76.4 ± 0.30	78.8 ± 0.29	81.2 ± 0.29	84.1 ± 0.24	89.5 ± 0.17	92.4 ± 0.17

[1] operationalised as actual activity being larger than baseline + 2 baseline standard-deviations

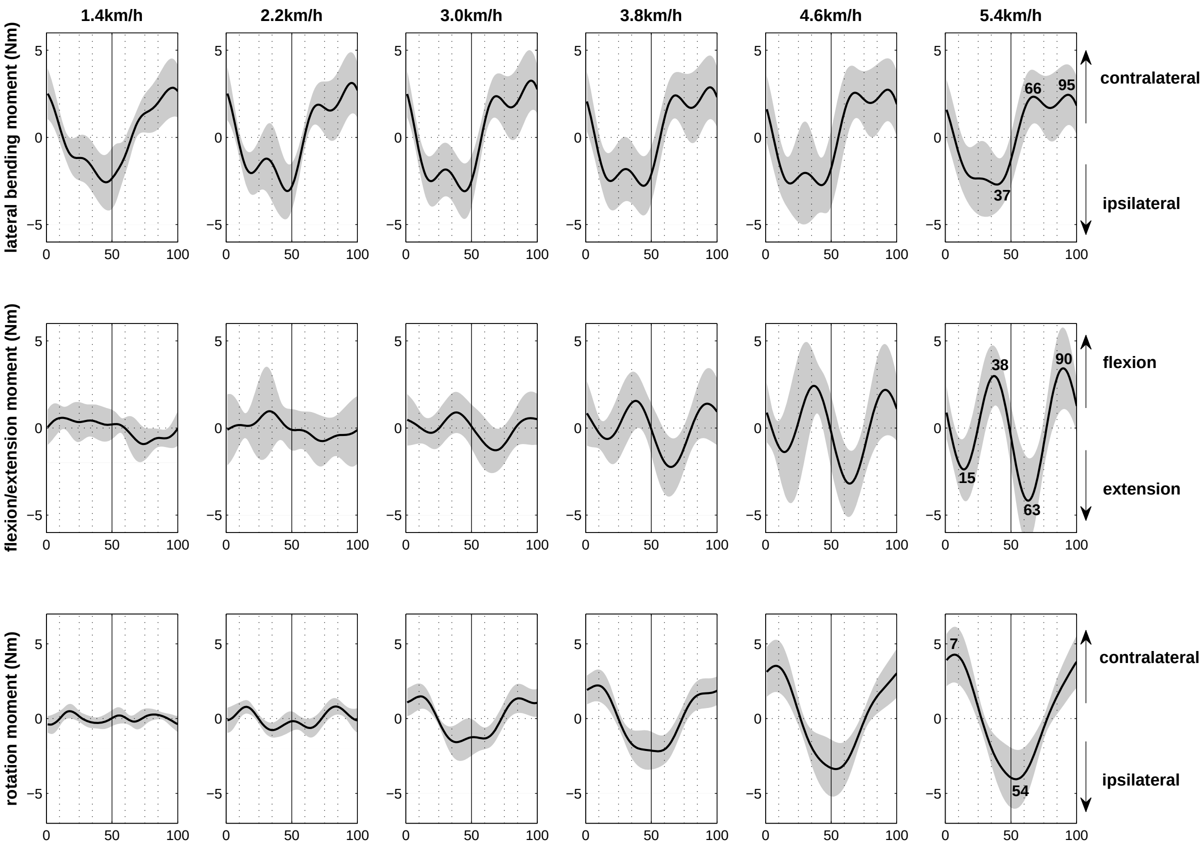

Fig. 1 Ensemble averaged trunk moments (Nm) as a function of stride cycle: lateral bending (first row), flexion/extension (second row), and trunk rotation (third row). Bold curves represent mean values over subjects, and shaded grey areas variability (± the between-subject standard-deviation at each time point). In the right panels, the average times (% of the stride cycle) of the peaks in the moment curves are given. Dotted vertical lines represent 10, 25, 35, 60, 75, and 85% of the stride cycle, the solid vertical lines represent 50%.

Due to artefacts in the EMG signal, TA data of one subject was discarded. For all other signals, raw EMG data were high-pass filtered at 250 Hz (1st order Butterworth) to remove ECG contamination and to obtain less variable estimates of EMG amplitude (Potvin & Brown 2004; Staudenmann, Potvin, Kingma, Stegeman & Van Dieën, 2007). Then, data were full-wave rectified, and low-pass filtered at 5 Hz (2nd order Butterworth) to obtain the linear envelope. Baseline activity, as recorded with the subject lying supine and relaxed, was subtracted from this linear envelope (Fig. 2). For the analysis of “percentage of time active”, a more conservative estimate was

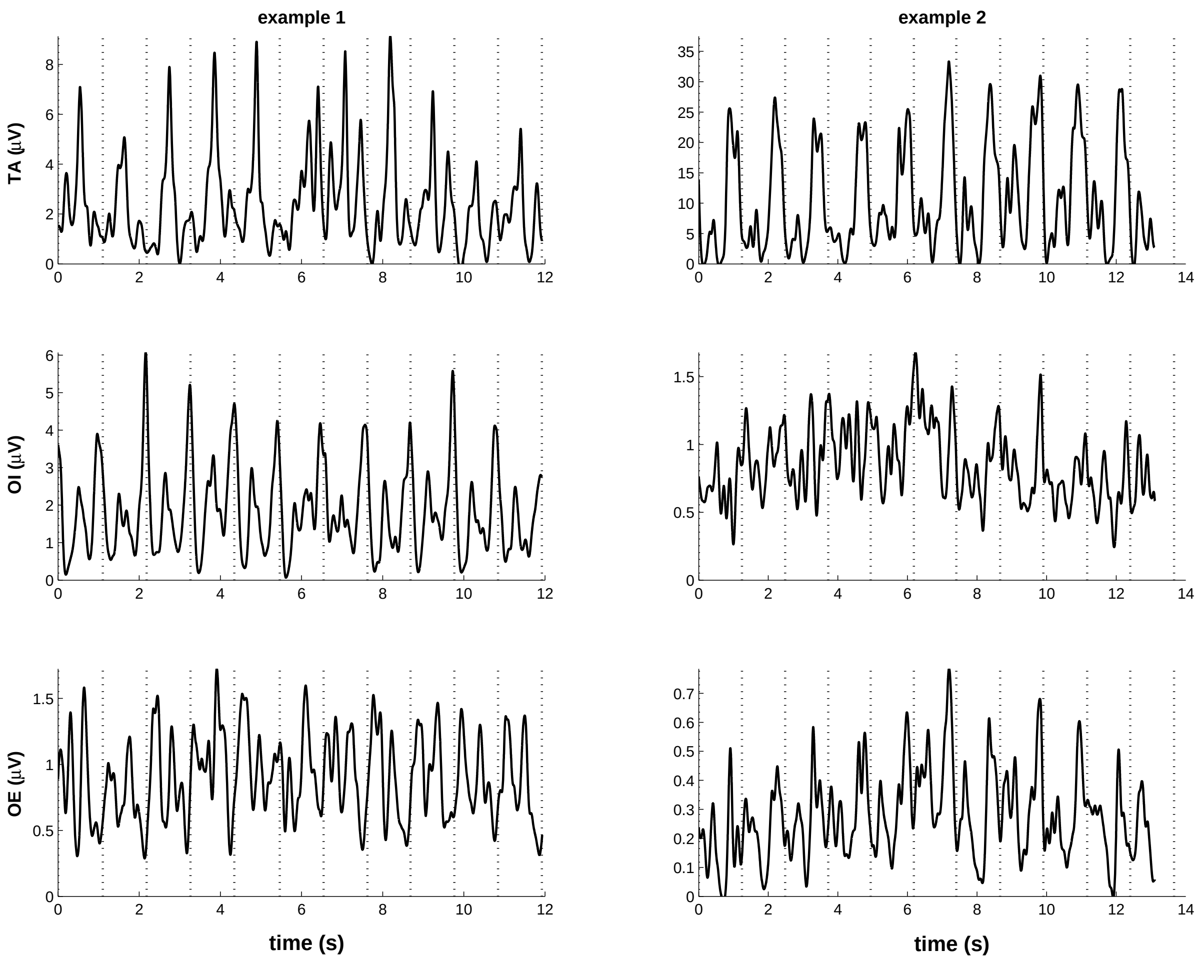

Fig. 2 Time series of EMG linear envelopes (μV) of TA, OI and OE, at preferred speed (3.8km/h). Dotted vertical lines represent ipsilateral heel contacts. Example 1 (left) was chosen from the subjects who had a triphasic pattern of TA activity at higher speeds; example 2 (right) is one of the two subjects who did not.

used, i.e., the percentage of measurement points with EMG activity above baseline plus two baseline standard-deviations (Table 2).

Similar as in the analysis of the moments, the time series of EMG linear envelopes were time-normalized, averaged over strides per time point per subject per speed, and then averaged over subjects (Fig. 3). For further graphical analysis, EMG amplitudes within the stride cycle were grouped per muscle per subject per speed, into three levels: lower than 10%, between 10% and 50%, and higher than 50%, of actual maximum activity. The distributions of these levels were plotted in the time domain per subject for 2.2 km/h, 3.8 km/h and 5.4 km/h (Fig. 4, upper three panels), and the averages over subjects were projected onto the phase map (Fig. 4, bottom).

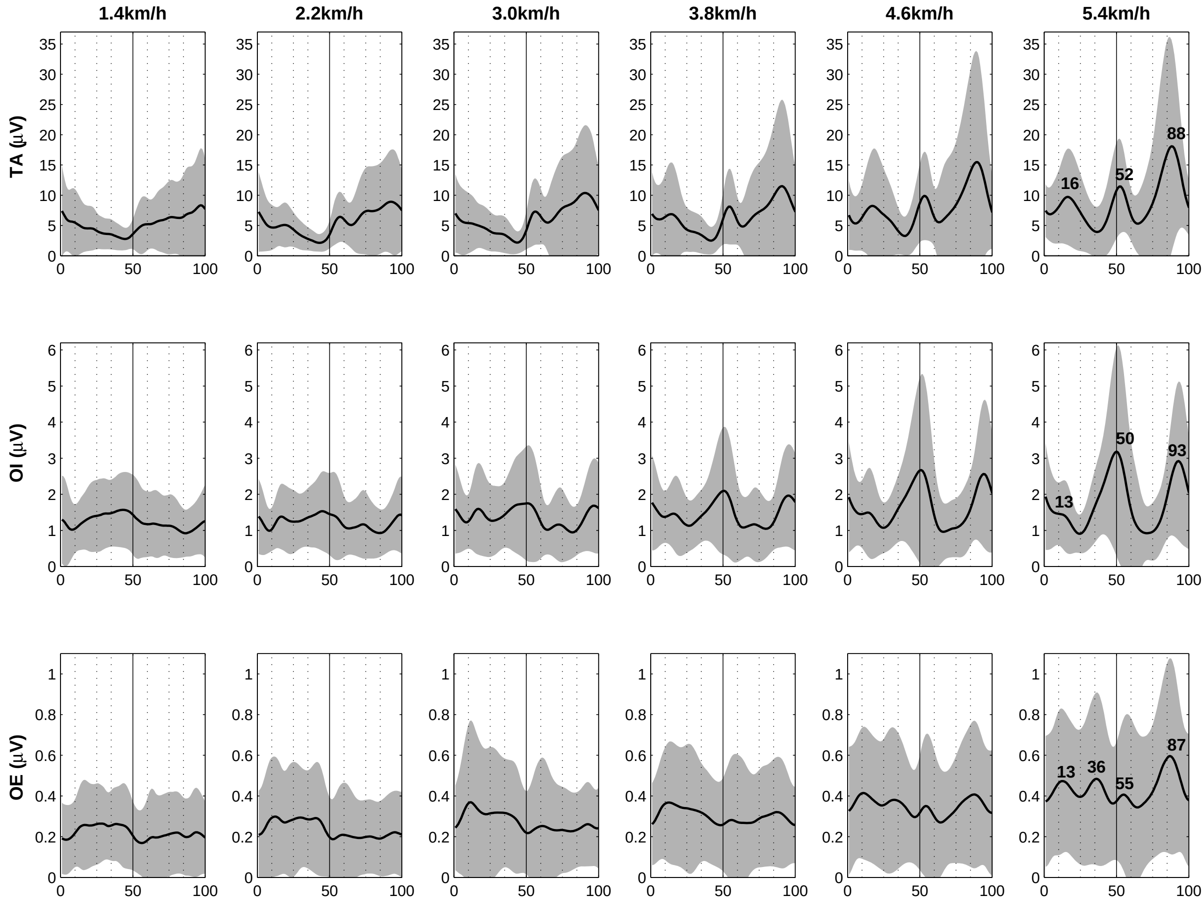

Fig. 3 Ensemble averaged EMG linear envelopes (µV) per muscle (TA, OI and OE) as a function of stride cycle. Bold curves represent mean values over subjects, and shaded grey areas variability (± the between-subject standard-deviation at each time point). In the right panels, the average peak times (% of the stride cycle) of EMG activity are given. Dotted vertical lines represent 10, 25, 35, 60, 75, and 85% of the stride cycle, the solid vertical lines represent 50%.

PCA, Fourier analysis, and cross correlations

To assess between-speed and between-subject consistency in the moment and in the EMG data, Principle Component Analysis (PCA) was used (Daffertshofer, Lamoth, Meijer & Beek, 2004). PCA decomposes a multivariate data set into successive weighted time series, “principle components”, mutually independent (“orthogonal”), and ordered along the amount of variance each contains.

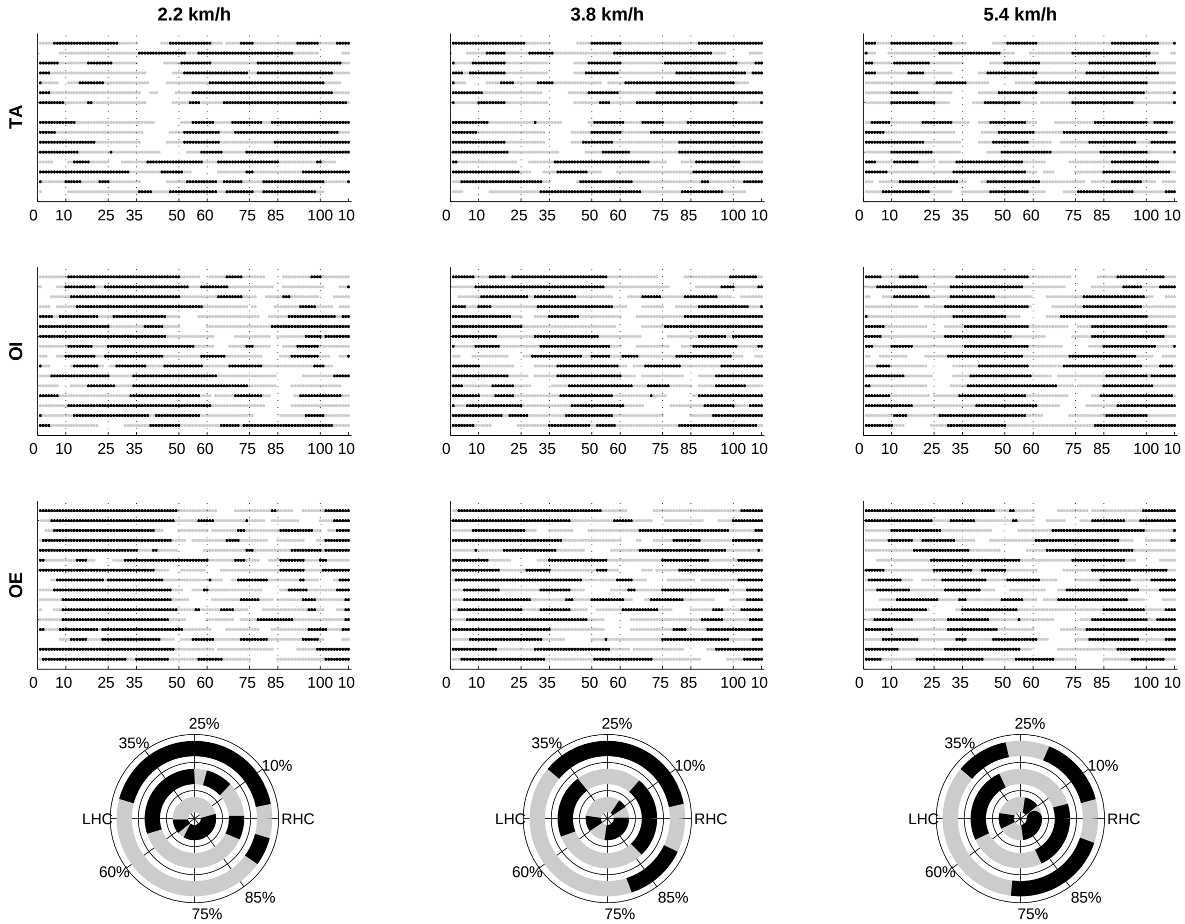

Fig. 4 Muscle activity at each time point of the stride cycle in each individual subject of the TA, OI, and OE (top three rows) at 3 different speed levels (columns). Black bars represent muscle activity above the 50th percentile amplitude for each subject per speed, while white bars represent muscle activity below the 10th percentile, and grey bars represent amplitudes in between. In the bottom row, these results have been averaged over subjects, and projected onto the sketch map of the stride cycle, using the same code of grey shades. TA activity is depicted in the inner ring, OE in the outer ring, and OI in between.

To identify the frequencies at which muscles were active, Fourier transformation of the linear envelopes was performed by using Welch's periodogram method (Hamming windows: 1000 samples, overlap: 750 samples). Peaks in the power spectrum were located, and absolute as well as relative power (as % of total power) was determined at each peak. Relative power per main frequency was graphically depicted per muscle at all speed levels (Fig. 5).

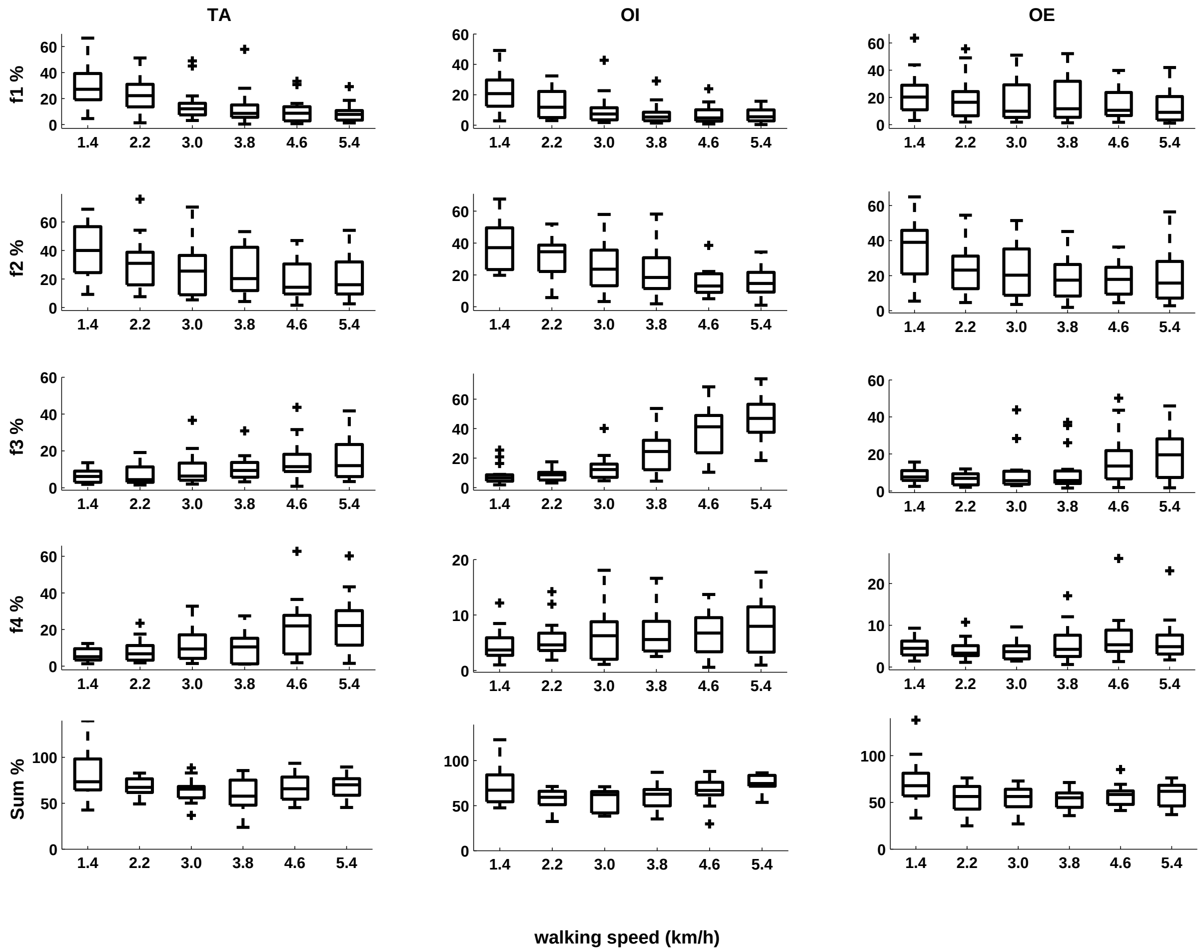

Fig. 5 Relative power as percentage of total power (vertical axis) at the four main frequencies at each speed, with f1 corresponding to normal breathing frequency, f2 to stride frequency, f3 twice the stride (= step) frequency, and f4 three times the stride frequency. The bottom row presents the total relative power of the four frequencies taken together. Each box runs from the 25- to the 75-percentile, the transverse line indicates the median, "+" represents outliers, and the error bars represent the range, excluding the outliers.

Correlations between trunk moments in the three directions were calculated to verify whether they were sufficiently independent to assess the relation of each moment with muscle activity. Subsequently, the peaks of the cross correlations between moments and EMG linear envelopes (at a short positive lead of the EMG signals, < 0.5 s, to account for electromechanical delay), were calculated. Depending on the mechanical function, inferred from anatomy, either the positive or the negative peak in the cross-correlation function was determined.

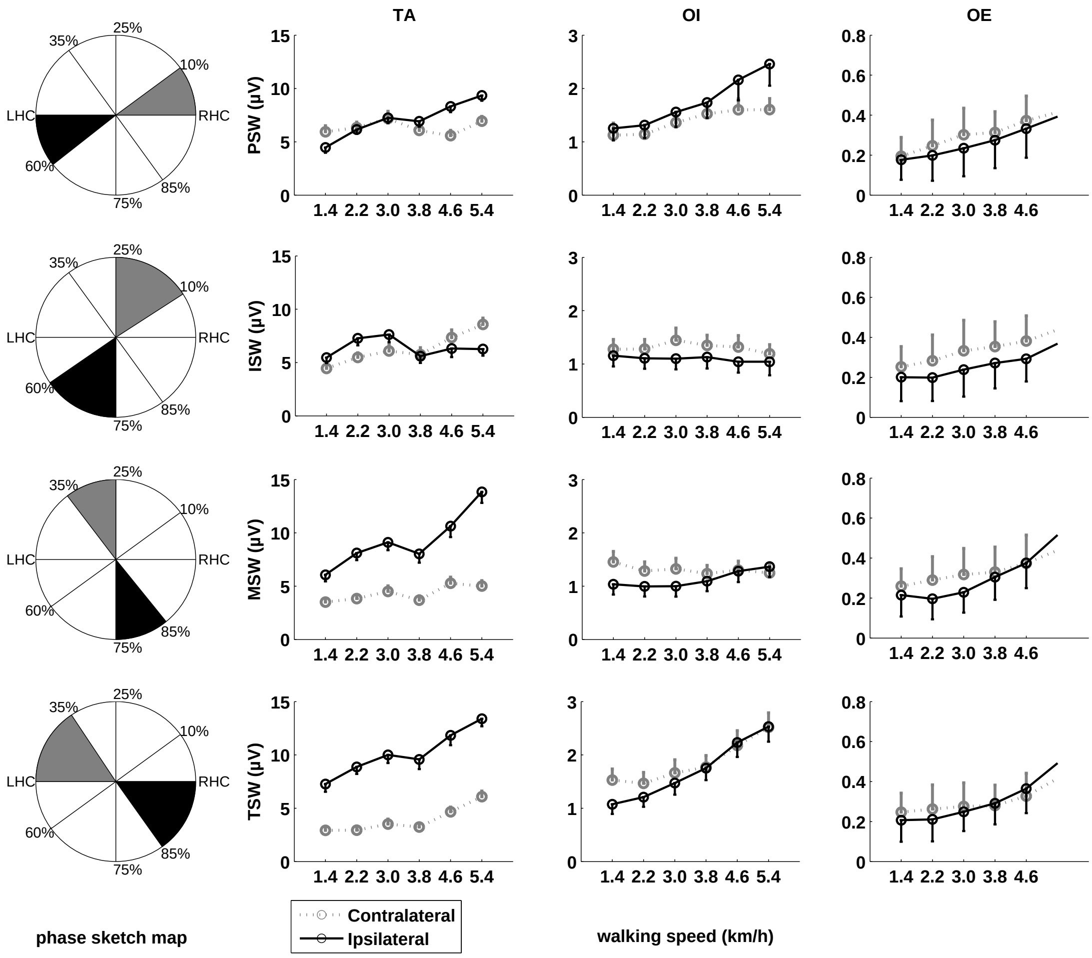

Fig. 6 Muscle activity (μV) during ipsilateral and contralateral swing phases, averaged over subjects at each speed. Error bars represent between-subject standard-errors. The left column specifies the ipsilateral (black) and contralateral (grey) swing phase for which the graphs are given per muscle.

GEE analyses

To analyse the within-stride modulation of trunk moments, General Estimation Equations (GEEs, cf. Liang & Zeger, 1993) were calculated for the average within-stride standard-deviation per subject, with Speed as covariate. Similar GEE analysis was performed on the absolute and relative power of the main EMG frequencies, again with Speed as covariate (Table 3). Since stabilisation requires a symmetrical component in muscle activity (Hodges, 1999, 2008; Hu et al., 2010), GEEs were also performed on mean EMG activity per muscle per phase, with Speed as covariate and Side as factor (Table 4; Fig. 6). Finally, to differentiate how each of the three

Table 3 Model *p*-values (*p*) and regression coefficients (*B*) from GEEs, with Speed as covariate, of the absolute and relative power of the four main frequencies in the EMG linear envelopes of the transversus abdominis (TA), obliquus internus (OI) and obliquus externus (OE).

Muscle	Frequency[1]	absolute power				relative power			
		Intercept		Speed[2]		Intercept		Speed	
		p	*B*	*p*	*B*	*p*	*B*	*p*	*B*
TA	f1	**0.00**	28.52	0.54	0.49	**0.00**	26.90	**0.00**	-4.03
	f2	0.19	60.99	**0.049**	16.53	**0.00**	36.08	**0.00**	-3.66
	f3	0.08	-11.74	**0.02**	14.76	**0.00**	5.73	**0.00**	2.06
	f4	0.08	-33.53	0.06	22.06	**0.00**	4.62	**0.00**	3.46
OI	f1	**0.03**	0.89	0.61	-0.02	**0.00**	17.71	**0.00**	-2.73
	f2	0.47	0.29	0.21	0.32	**0.00**	35.86	**0.00**	-4.56
	f3	**0.02**	-2.82	**0.01**	1.63	0.06	2.80	**0.00**	8.01
	f4	0.31	-0.27	0.09	0.26	**0.00**	4.89	**0.04**	0.58
OE	f1	0.24	0.02	**0.01**	0.01	**0.00**	21.82	**0.00**	-1.59
	f2	0.89	0.004	0.13	0.02	**0.00**	29.88	**0.00**	-2.87
	f3	0.33	-0.02	**0.01**	0.02	**0.00**	5.66	**0.00**	2.55
	f4	**0.04**	-0.01	**0.01**	0.01	**0.00**	3.94	**0.02**	0.54

[1] f1: breathing, f2: stride, f3: step, f4: three times stride.
[2] speeds (from 1.4 km/h to 5.4 km/h) were coded 0-5.

muscles was involved in the movements in the three planes, GEEs on peak cross-correlations between moments and EMG activity were calculated with Speed as covariate and Muscle as factor (Table 5). Note that muscle activities per se cannot be compared, whereas these cross-correlations can.

All GEE analysis was performed with SPSS 16, with significance set at $p < 0.05$.

Results

Stride frequency and step length

All subjects walked at all speeds. Stride frequency increased with speed from 0.52 Hz to 0.99 Hz, and step length from 0.38 m to 0.76 m (Table 1). Step length was largely symmetric, with an average difference between right and left of 0.005 ± 0.03 m.

Table 4 Model *p*-values (*p*, bold when significant) and regression coefficients (*B*) from GEEs, with Side as factor (ipsilateral versus contralateral swing) and Speed as covariate, on mean EMG amplitudes in the different phases of the stride cycle of the (right-sided) transversus abdominis (TA), obliquus internus (OI) and obliquus externus (OE). Non-significant interactions were omitted from the models.

Muscle	*N*	Phase[1]	Intercept		Side		Speed[3]		Side × Speed	
			p	*B*	*p*	*B*[2]	p	*B*	*p*	*B*[2]
TA	15	PSw	**0.00**	5.68	0.07	-1.08	0.28	0.19	**0.00**	0.81
		ISw	**0.00**	3.94	**0.01**	1.86	**0.01**	0.87	**0.01**	-0.74
		MSw	**0.00**	3.21	**0.01**	2.78	0.06	0.40	**0.02**	0.88
		TSw	**0.00**	1.41	**0.00**	6.36	**0.00**	0.98		
OI	16	PSw	**0.00**	0.94	0.14	0.35	**0.00**	0.18		
		ISw	**0.00**	1.35	0.054	-0.21	0.60	-0.02		
		MSw	**0.00**	1.25	0.13	-0.18	0.47	0.02		
		TSw	**0.00**	1.34	**0.00**	-0.39	**0.00**	0.21	**0.02**	0.10
OE	16	PSw	**0.00**	0.20	**0.03**	-0.04	**0.00**	0.04		
		ISw	**0.00**	0.25	**0.00**	-0.08	**0.00**	0.03		
		MSw	**0.00**	0.22	0.37	-0.03	**0.00**	0.05		
		TSw	**0.00**	0.23	**0.00**	-0.06	**0.00**	0.03	**0.01**	0.03

1 PSw: Pre-swing, ISw: Initial swing, MSw: Mid-swing, TSw: Terminal swing.

2 Regression coefficient during ipsilateral swing.

3 Speeds (from 1.4 km/h to 5.4 km/h) were coded 0-5.

Note that GEEs calculate regression equations, and, for instance, the first line reads as: the amplitude of TA activity in PSw = 5.68 - 1.08 (during ipsilateral swing) + 0.19 × Speed + 0.81 × Speed (during ipsilateral swing).

Table 5 Cross-correlations (averaged across subjects ± standard-deviations) between trunk moments (LA=lateral bending, FL = flexion/extension, RO = rotation) and the EMG linear envelopes of the transversus abdominis (TA), obliquus internus (OI) and obliquus externus (OE).

Speed	1.4 km/h	2.2 km/h	3.0 km/h	3.8 km/h	4.6 km/h	5.4 km/h
Trunk Lateral bending vs. Muscle activity						
LA *vs.* TA	0.40 ± 0.18	0.41 ± 0.22	0.43 ± 0.18	0.44 ± 0.18	0.40 ± 0.18	0.43 ± 0.20
LA *vs.* OI	0.39 ± 0.17	0.38 ± 0.18	0.37 ± 0.14	0.35 ± 0.15	0.39 ± 0.16	0.48 ± 0.12
LA *vs.* OE	0.23 ± 0.21	0.32 ± 0.21	0.34 ± 0.15	0.30 ± 0.12	0.33 ± 0.12	0.34 ± 0.12
Trunk Flexion/Extension vs. Muscle activity						
FL *vs.* TA	0.18 ± 0.11	0.22 ± 0.11	0.28 ± 0.18	0.31 ± 0.16	0.33 ± 0.11	0.38 ± 0.18
FL *vs.* OI	0.18 ± 0.13	0.24 ± 0.11	0.38 ± 0.13	0.38 ± 0.19	0.48 ± 0.14	0.61 ± 0.12
FL *vs.* OE	0.19 ± 0.09	0.23 ± 0.13	0.30 ± 0.17	0.36 ± 0.15	0.36 ± 0.18	0.40 ± 0.23
Trunk Rotation vs. Muscle activity						
RO *vs.* TA	0.16 ± 0.09	0.31 ± 0.12	0.38 ± 0.17	0.43 ± 0.16	0.42 ± 0.15	0.42 ± 0.19
RO *vs.* OI	0.26 ± 0.09	0.28 ± 0.09	0.40 ± 0.16	0.40 ± 0.13	0.43 ± 0.15	0.46 ± 0.14
RO *vs.* OE	0.19 ± 0.11	0.32 ± 0.14	0.42 ± 0.17	0.41 ± 0.15	0.41 ± 0.17	0.40 ± 0.20

Moments

To assess the consistency of trunk moment data, PCA was used over all subjects and all speeds. The first principal component contained 80.4% (lateral bending), 56.2% (flexion/extension), and 90.1% (rotation) of the variance. These numbers are rather high, implying considerable between-subject and between-speed consistency of the trunk moments, though somewhat less in the sagittal plane. Still, GEE analysis revealed that the within-stride standard-deviation (the "modulation") of the moment over the stride cycle (cf. Fig. 1) increased with speed in flexion/extension (p = 0.03) and rotation (p = 0.003), but not in lateral bending (p = 0.98).

At the highest speed (Fig. 1, right panels), the average lateral bending moment had a peak towards ipsilateral in contralateral terminal swing (37% of the stride cycle), and towards contralateral in ipsilateral mid-swing (66%) and terminal swing (95%). The flexion moment peaked in terminal swing (38% and 90%), and extension in pre-swing (15% and 63%). The contralateral rotation moment peaked in contralateral pre-swing (7%), and ipsilateral rotation in ipsilateral pre-swing (54%). In the Discussion, these findings will be related to peaks in EMG activity.

EMG

In the literature, it was suggested that tonic muscle activity may be indicative of muscle function (Saunders et al., 2004). In the present study, all three lateral abdominal muscles appeared to be active most of, or even all the time (cf. Fig. 2). With strict operationalization of "activity", i.e., more than two standard-deviations above baseline, the percentage of time active ranged from 91-96% for TA, 97-98% for OI, and 76-92% for OE (Table 2).

Fig. 3 gives average EMG linear envelopes over the stride cycle, averaged over subjects. PCA analysis over speeds and subjects revealed that the first mode contained 64.7% of the variance in TA, 58.0% in OI, and 46.9% in OE. These percentages are somewhat lower than those of trunk moments. In PCAs of EMG per subject over speeds, the first mode contained on average more variance: 78.5% in TA, 79.9% in OI, and 73.6% in OE. Clearly, EMG activity of the lateral abdominal muscles was more consistent within subjects across speeds than between subjects.

At the highest speed (Fig. 3, right panels), TA peaked in contralateral initial swing (16% of the stride cycle), ipsilateral pre-swing (52%), and ipsilateral terminal swing (88%). OI presented a hump in contralateral initial swing (13%), and peaks at contralateral heel contact (50%) and ipsilateral terminal swing (93%). The three major peaks of OE activity occurred in contralateral initial swing (13%), contralateral terminal swing (36%), and ipsilateral terminal swing (87%). As indicated before, these data will be related to the corresponding moment data in the Discussion.

Power spectrum analysis of the EMG linear envelopes identified four common frequency components, which suggests that the muscles were involved in multitasking. EMG modulation at the lowest frequency (f1) was likely related to breathing. We did not measure breathing frequency itself, but f1 was close to 0.3 Hz, the normal breathing frequency during gait (White & McNair, 2002). The second peak (f2), 0.5-1 Hz, corresponded with stride frequency, the third (f3) with twice this frequency, i.e., step frequency, and the fourth (f4) was found at three times stride frequency. Fig. 4 presents a more detailed illustration of muscle activity at stride frequency (i.e., once per cycle, such as TA activity during ipsilateral swing at 2.2 km/h), and step frequency (i.e., bilateral symmetry, such as OI activity at and around terminal swing at 5.4 km/h).

The relative power of all four main frequencies taken together hovered around 60% of total power (Fig. 5, bottom row). GEE analyses of relative power consistently revealed significant effects of speed (Table 3, right half; Fig. 5, upper rows), with relative power at f1 and f2 decreasing, and at f3 and f4 increasing with increasing speed. For absolute power, the pattern was different (Table 3, left half). Whenever the effect of speed was significant (that is, in TA at f2 and f3, OI at f3, and OE at f1, f3, and f4), the regression coefficient was positive, implying an increase of absolute power with speed.

GEEs of mean EMG activity per muscle per phase, with Speed as covariate and Side (ipsilateral vs. contralateral swing) as factor (Table 4), revealed significant effects of Side in TA (but not in pre-swing), with more ipsilateral than contralateral activity. Except in terminal swing, OI activity was not significantly different between sides. OE was mostly active during contralateral swing (except in mid-swing). Hence, OI activity was largely symmetrical, while TA and OE activity were not.

Relations between moments and EMG

Between-moment correlations ranged from 0.0 to 0.2, indicating that the moments were sufficiently independent to separately assess the relation of each moment with muscle activity. Peak cross-correlations between moment and EMG activity (Table 5) were around 0.4 at higher speeds, except for the correlation between the flexion/extension moment and OI activity, which increased to approximately 0.6. Overall, these results implied that all three lateral abdominal muscles were consistently involved in movements in all three planes.

In GEEs with Muscle as factor and Speed as covariate, lateral bending moments had higher cross-correlations with TA and OI than OE activity (TA vs. OI: $p = 0.57$; TA vs. OE: $p = 0.02$; OI vs. OE: $p = 0.01$), without a clear effect of speed ($p = 0.053$). For flexion/extension moments, cross-correlations were higher with OI than TA and

OE (OI vs. TA: $p = 0.05$; OI vs. OE: $p = 0.03$; TA vs. OE: $p = 0.99$), and increased significantly with speed ($p = 0.00$). Cross-correlations between the rotation moments and EMG were similar for all three muscles (OI vs. TA: $p = 0.60$; OI vs. OE: $p = 0.72$; TA vs. OE: $p = 0.93$), and also increased with speed ($p = 0.00$). In sum, lateral bending revealed most involvement of TA and OI, and flexion/extension of OI.

Discussion

The effects of speed

We expected that the same functions would be performed by the lateral abdominal muscles at all speeds, and that no function would be ignored at the higher speeds. The necessity to control trunk motion in three planes defines mechanical constraints for the control of the lateral abdominal muscles. In the frontal plane, trunk muscles need to maintain equilibrium against the moment caused by gravity during single stance. These lateral bending moments thus varied with stride frequency, and were relatively constant in magnitude over gait speeds. In the sagittal plane, moments perturbing trunk equilibrium are probably reaction moments related to hip moments, requiring internal extension moments in early swing and flexion moments in late swing (cf. Fig. 1), both becoming more important at higher speeds. Finally, in the transverse plane, trunk muscle activity controls stride-related rotations around the longitudinal axis. Again, these moments increased with speed, reflecting the increase in trunk rotation amplitude that occurs with speed (Bruijn, Meijer, Van Dieën, Kingma & Lamoth, 2008).

PCA of trunk moments revealed a high consistency of the moments in the frontal and transverse planes over subjects and speeds, with 80% or more of the variance contained in the first principal component. In the sagittal plane, consistency was lower (56% of variance in the first mode), probably due to low frequency fluctuations of these moments reflecting changes in trunk inclination. Overall, all three mechanical constraints applied in a similar way at all speeds, notwithstanding the fact that flexion/extension and rotation moments scaled with walking speed.

EMG linear envelopes were less consistent than the moments, in particular between subjects. Nevertheless, the first principle components contained more than 70% of the variance within subjects (cf. Ivanenko et al., 2008). Relatively low consistency of EMG envelopes is partially due to the inherent variability of the EMG signal, which does not reflect variability of muscle activation (Staudenmann, Roeleveld, Stegeman & Van Dieën, 2010). Inspection of Fig. 3 suggests that patterns of muscle activity changed considerably with speed (Anders et al., 2007; Ivanenko, Poppele & Lacquaniti, 2006), but this appears to result from quantitative changes (e.g., scaling, and some phase shifts), rather than the emergence of a qualitatively different pattern (cf. Ivanenko et al., 2008; Vasudevan & Bastian, 2010). Thus, the consistency of the EMG patterns across speeds, and the fact that relative power at the four main frequencies changed gradually over speed (Fig. 5), supported the hypothesis that the same constraints can be found in EMG activity at all speeds, albeit with different weights. Note that the relative power at breathing frequency (f1) decreased substantially with speed (as in Saunders et al., 2004), while absolute power did not (Table 3). Also, the decrease in relative (but not absolute) power at f2 may be misleading, as this component is also present in f4 at the higher speeds.

Clearly, some functions of the lateral abdominal muscle gain importance with increasing walking speed, but no function is ignored to deal with these constraints.

Tonic activity

The lateral abdominal muscles were active throughout most of the stride cycle at all speeds. This tonic EMG activity is likely related to the need for co-contraction to stabilize the spine around the neutral posture (Cholewicki, Panjabi & Khachatryan, 1997). Moreover, previous research indicated that bilateral contractions of the lateral abdominal muscles stabilize the sacroiliac joints (Hu et al., 2010).

Note that "tonic" in this context implies continuous activity, while not excluding phasic modulation. Saunders et al. (2004) measured fine-wire EMG of the lateral abdominal muscles during gait at 1 m/s. OI was tonically active in three of the seven, but TA in all seven subjects. It was concluded that tonic activity marks a difference between TA and the other abdominal muscles. This, however, was not confirmed in the present study, as all three lateral abdominal muscles were active above baseline most of the time. Note that the use of surface EMG for OI and OE in the present study may have affected these results. Crosstalk of OI in the OE signal and of TA in the OI signal cannot be excluded with surface EMG. On the other hand, fine-wire EMG may yield an unrepresentative sample from these large and heterogeneous muscles (cf. Urguhart & Hodges, 2005).

EMG modulation at different frequencies

EMG modulation at the lowest frequency (f1) was likely related to breathing. Previous studies also related the activity of all three lateral abdominal muscles to respiration (Beales, O'Sullivan & Briffa, 2009a; Hodges et al., 2002; Saunders et al., 2004). At the lowest speed, respiration took around 20% of all power, which decreased to less than 10% at the highest speeds, apparently because other functions then required more power. Note, again, that the absolute power of f1 did not decrease with increasing speed (Table 3).

Modulation at the stride frequency (f2) was most clearly visible in the time series at the lower speeds, while higher frequency modulations were still limited in magnitude (Table 3, Fig. 4). The f2 frequency component derives from unilateral muscle activity, probably related to trunk lateral bending and rotation. At the highest speed (Fig. 3, right panels), there were peaks of TA and OE activity in contralateral initial swing, coinciding with a hump in the curve of OI. This muscle activity appears to be related to lateral bending, balancing the pull of gravity when the contralateral leg is the swing leg. The other f2 peaks of TA and OI occurred at contralateral heel contact, approximately simultaneous with the peak in the ipsilateral rotation moment (Fig. 1), suggesting that TA and OI acted as ipsilateral rotators (Benninghoff & Goerttler, 1964; Kumar, Narayan & Garand, 2003; Hodges, 2008).

Modulation of EMG amplitudes at step frequency (f3) indicates bilateral muscle activity, clearly visible in the time series at the highest speed (Fig. 4, right panels). Such activity of OI and OE is probably related to the modulation of trunk flexion moments (Fig. 1, right middle panel). Moreover, symmetric bilateral activity of the lateral abdominal muscles has been associated with spinal (Hodges, 1999, 2008) and sacroiliac (Hu et al., 2010) stabilization. The relative power at f3 increased with gait speed, as did the amplitude of the flexion/extension moment (cf. Fig. 1). Also stabilization demands will increase with increasing speed. At the highest speed, peak muscle activity was found just before ipsilateral heel contact (Fig. 3, right panels) in all three muscles, and OI revealed major activity around contralateral heel contact, while there was a peak of TA and a small peak of OE just after contralateral heel contact. Part of OI activity at contralateral heel contact is probably related to ipsilateral rotation (f2), but this large peak (Fig. 3, right middle panel) will also contribute to flexion (f3). In general, bilateral muscle activity peaked at or before heel contact, which may be a destabilizing event for the lumbar spine and the sacroiliac joints. TA has been shown to be bilaterally activated in anticipation of different trunk perturbations (Cresswell, Oddsson & Thorstensson, 1994; Hodges & Richardson, 1997a, 1997b; McCook, Vicenzino & Hodges, 2009). Nevertheless, in the present study, the activity was more symmetric in OI than in TA (Figs. 4 & 6), suggesting a predominant role for OI in stabilizing the lumbar spine and pelvis during gait, perhaps because of its simultaneous contribution to the required flexion moments at terminal swing (McGill, 1996).

The fourth frequency (f4) equalled three times the stride frequency. This triphasic pattern appears to have emerged from the summation of f2 and f3 components, that is to say, simultaneous engagement in stride-related and step-related activities. Note that the relatively high power of f4 in TA at higher speeds partly reflects that TA had considerable bilateral activity at the higher speeds.

In sum, all three muscles contribute to multiple functions in gait, both consecutively (task switching) and simultaneously (multitasking), and no single muscle is specifically dedicated to a single task. All three muscles were active most or all of the time, presumably to stabilize the spine and pelvis. Moreover, the lateral abdominal muscles were engaged in breathing, and were active at the frequency of stride-related constraints, probably to control trunk movement in the frontal and transverse planes, as well as the frequency of step-related constraints, likely to control trunk orientation in the sagittal plane. The muscles also appeared to deal with potentially destabilizing events such as heel contact.

Conflicting constraints

Conflicting constraints are illustrated by comparing peak muscle activity (Fig. 3) with peak trunk moments (Fig. 1) at the highest speed (right panels in both Figs.).

At contralateral initial swing, all three muscles appear to be involved in lateral bending, with a peak for TA at 16%, a hump in the OI curve at 13%, and peaks for OE activity at 13% and 36% of the stride cycle. Correspondingly, the moment curve (Fig. 1, right upper panel) reveals an ipsilateral bending moment from the end of contralateral pre-swing, with a peak at 37% of the stride cycle. The above peaks in

TA and OI activity would contribute to ipsilateral rotation, while the rotation moment is still toward the contralateral side, switching to ipsilateral rotation around 25% (Fig. 1, right lower panel). This "unwanted" ipsilateral rotation moment of TA and OI appears to be offset by simultaneous OE activity, i.e., co-contraction of an antagonist.

The activity of all three muscles in contralateral initial swing will exert a flexion moment in the sagittal plane. However, it is known (e.g., Lamoth et al., 2004; cf. Saunders et al., 2004; Ceccato et al., 2009) that the lumbar erector spinae is bilaterally active from just before heel contact into initial swing. The net moment in contralateral initial swing shows an extension peak at 15% of the stride cycles. Apparently, the extension moment caused by the lumbar erector spinae offsets the flexion moment caused by (co-contraction of) the lateral abdominal muscles.

Around contralateral heel contact at the higher speeds, there are peaks in TA activity (cf. Fig. 3, right upper panel, 52%) and OI activity (right middle panel, 50%), but no, or hardly any, corresponding peak of OE. The bursts of TA and OI can be related to the ipsilateral rotation moment, with a peak at 54% of the stride cycle (Fig. 1, right lower panel), while OE is a contralateral rotator. TA and OI will also exert a flexion moment; still, at contralateral contact (50%), the internal flexion/extension moment is close to zero, and lumbar erector spinae activity (Lamoth et al., 2004; cf. Saunders et al., 2004; Ceccato et al., 2009) appears to balance the activity of abdominal flexors, which is another example of dealing with conflicting constraints through co-contraction of antagonists.

Muscle activity at contralateral heel contact not only belongs to f2, but is also part of f3, with similar peaks close to ipsilateral heel contact. It is unclear why peaks occur somewhat earlier before ipsilateral (88, 93, and 87%, Fig. 3, right panels) than before contralateral heel contact (52, 50, and 55%). Considering the approximate symmetry of these peaks, they might be associated with the potentially destabilizing effect of heel contact and weight acceptance on the pelvis and spine. The peaks in TA (88%) and OI (93%) coincided with a rotation moment to the contralateral side (increasing towards its peak at 7%, Fig. 1, right lower panel), and their ipsilateral rotation is thus an "unwanted" effect, which appears more than compensated by the peak (87%, Fig. 3, right lower panel) of OE. We only measured EMG in right abdominal muscles, but given that activity around heel contacts is largely symmetrical, similar probably happens with left TA, OI, and OE.

As hypothesized, conflicting constraints require coordinated activity of the lateral abdominal muscles (and the lumbar erector spinae), and at several points in the gait cycle co-contraction of antagonists is used to offset unwanted mechanical effects of individual muscles.

Limitations

It can be argued that speed was experimentally manipulated in the present study. Nevertheless, the study was largely descriptive, using biomechanics to understand muscle coordination in multitasking. Each specific explanation still awaits the test of experiments. The study was limited to young healthy women, whereas gender (Anders, Wagner, Puta, Grassme & Scholle, 2009) as well as pathology (e.g., Huang et al., 2011) are known to affect gait.

Conclusion

During gait, all three lateral abdominal muscles are involved in simultaneous task execution (multitasking) as well as consecutive task execution (task switching). Activity of the lateral abdominal muscles contributes to breathing, trunk motion in three dimensions, and lumbar spine as well as pelvic stability. Task execution is distributed over all three muscles, and no muscle is exclusively assigned a particular task. The effects of speed are gradual, without sudden transitions, and no constraint is ignored at the higher speeds. Most important, the motor control system deals with conflicting constraints by coordinating the activities of the lateral abdominal muscles (and the lumbar erector spinae), with often several muscles being active simultaneously. In cases of conflicting constraints, the coordination of muscle activity appears to consist of co-contraction of antagonists to offset unwanted mechanical effects of agonists.

Chapter 5

Symmetry and asymmetry of abdominal wall muscle activity during the Active Straight Leg Raise (ASLR)

Hu H, Meijer OG, Van Dieen JH, Hodges PW, Bruijn SM, Strijers RL, Nanayakkara PW, Van Royen BJ, Wu W, Xia C. Manual Therapy, submitted.

Abstract

The Active Straight Leg Raise (ASLR) is a diagnostic test in pelvic girdle pain, hip pain, and low back pain. It is thought to test stabilization of the pelvis through “force closure”, with transverse and oblique abdominal muscles pressing the iliac bones against the sacrum. This compression requires symmetric activation, but the abdominal muscles are engaged in multitasking, with symmetric and asymmetric task components. We studied symmetry and asymmetry of the activity of abdominal muscles, rectus femoris and biceps femoris, in 16 healthy women performing the ASLR, in a “normal” condition, with weight added above the ankle, and with a pelvic belt. Fine-wire electromyography (EMG) was used for the transversus abdominis, and all other muscles were recorded with surface EMG. All abdominal muscles were bilaterally active. Asymmetric abdominal muscle activity was higher with weight, and there appeared to be less symmetric activity with the pelvic belt. Activity of the obliquus internus and the transversus abdominis depended on the side of the ASLR, the latter muscle being most asymmetrical, especially when weight was added. Inter-individual variability was large. During the ASLR, hip flexors pull the ilium forward, which must be countered by contralateral hip extension and/or activity of the abdominal muscles. Thus, symmetrical force closure must ensure that the pelvis moves as one unit in the sagittal plane. Asymmetrical task components are superimposed to maintain the orientation of the pelvis. To better understand ASLR, future research should include the 3D kinematics and kinetics of the pelvis and trunk.

Introduction

Pelvic girdle pain affects over 20% of pregnant women (Wu et al., 2004a), can be part of chronic groin pain in athletes (Verrall et. al., 2001), or may result from trauma (cf. Kanakaris et al., 2011). There has been a recent increase in interest in pelvic girdle pain (e.g., Mulholland, 2005; Vleeming et al., 2008; Robinson et al., 2010; Gutke et al., 2010; Vermani et al., 2010). Several diagnostic examinations are commonly applied, especially the Active Straight Leg Raise (ASLR) test (Mens et al., 2001, 2002). The ASLR is also used in the differential diagnosis of hip and groin pain (Cowan et al., 2004; Mens et al., 2006a), or lumbar disorders (Roussel et al., 2007). It is considered to be a clinical test of lumbopelvic stability, and has been suggested to assess the ability to effectively transfer load between the spine and legs via the pelvis (Mens et al., 1999, 2001). This ability is not only dependent on the passive properties of the sacroiliac joints and the symphysis, but also on motor control, more specifically, the activation of stabilizing muscles (De Groot et al., 2008; Beales et al., 2009a, b, 2010a, b; Hu at al., 2010, 2011; Jansen et al., 2010b).

Snijders proposed that bilateral activity of the transverse and oblique abdominal muscles can stabilize the pelvis, i.e., “force closure” (Snijders et al., 1993a & b; Vleeming et al., 1990a & b). In a recent study (Hu et al., 2010), we confirmed this idea for the ASLR, with less activity of the transverse and oblique abdominal muscles in a condition with a pelvic belt, which (partially) substitutes the force required. Force closure implies symmetrical muscle activity. Symmetrical increase of the thickness of the transversus abdominis (TA) and obliquus internus (OI) has been reported for healthy subjects drawing-in the abdominal wall (Hides et al., 2007), and subjects with unilateral pelvic girdle pain performing the ASLR (Teyhen et al., 2009). However, the symptoms of pelvic girdle pain have been found to correlate with asymmetric laxity of the sacroiliac joints (Buyruk et al., 1999; Damen et al., 2001, 2002), the pubic bones move asymmetrically in patients (Mens et al., 1999), and Beales et al. (2009b) reported symmetric activity of the obliquus externus (OE) and OI during an ASLR on the affected side, but asymmetrical activity during an ASLR on the nonaffected side. This pattern of results may be confusing, but we argue that the whole point is moot (cf. Hodges, 2008 vs. Allison et al., 2008). The ASLR is a mainly unilateral task, while force closure requires bilateral, symmetric activity. Actual muscle activity, therefore, must result from the interplay between asymmetric and symmetric task components. A more in-depth understanding of task components appears required to understand this interplay.

The present study aimed to better understand (a)symmetry of abdominal muscle activity during the ASLR in healthy subjects. Recent studies of abdominal muscle activity during gait show evidence of multitasking in the abdominal muscles (Saunders et al., 2004; Hu et al., submitted), with muscles engaged in several tasks simultaneously. In the present study, we expected to find similar multitasking. Assuming that force closure requires symmetric muscle activity, with other important components of the ASLR being asymmetrical, we expected that the actual pattern of symmetry and asymmetry in the activities of the muscles would result from the summation of the different task components. We used three conditions, one “normal”, one with a pelvic belt to reduce the symmetrical task component, and one with weight added above the ankle to increase the asymmetrical task component. In general, it will be rare for a muscle to exactly satisfy task constraints, and muscle activity will usually have unwanted side effects. In our abovementioned study on gait, we found that the control system deals with such conflicting constraints by simultaneously activating several muscles, with unwanted side effects of one muscle countered by the effects of another. We expected this also to be the case in the present study of abdominal muscle activity during the ASLR.

Considerable variability of muscle activation patterns between subjects performing the ASLR has been reported (e.g., Beales et al., 2010b). Between-subject variability was an additional focus of the present study.

Methods

Subjects

Sixteen healthy nulliparous females were enrolled (mean ± SD age 27.5 ± 2.7 years, weight 61.2 ± 9.8 kg, height 167.9 ± 7.6 cm, BMI 21.6 ± 2.4 kg/m^2). Exclusion criteria were: previous orthopedic surgery, walking-related disorders, or a history of low blood pressure. Participants gave written informed consent. The protocol was approved by the local Medical Ethical Committee.

Electromyography (EMG)

EMG was measured on the right side only. TA was recorded with intramuscular fine-wire electrodes (CE-marked paired hook-wire, 40 gauge insulated stainless steel, VIASYS Healthcare, Madison WI, USA), threaded into sterile 50 mm hypodermic needles, and trimmed, with 2-3 mm long "hooks" extending from the tip. After disinfection, the needle was inserted in semi-sterile conditions under ultrasound guidance. The insertion point was 2 cm medial to the midpoint of the vertical from the spina iliaca anterior superior to the rib cage (Hodges and Richardson 1997). Some subjects felt anxious and uncomfortable when the needle entered the muscle, but no lasting pain was reported. For OI, OE, rectus abdominis (RA), rectus femoris (RF), and biceps femoris (BF), EMG was recorded with pairs of surface electrodes (24 mm diameter Ag/AgCl discs, inter-electrode distance 20 mm; Kendall ARBO, Neustadt am Dom, Germany), following SENIAM recommendations (Hermens et al. 1999). Data (cf. Fig. 1, upper rows) were recorded at a sample frequency of 2000 samples/s with a multichannel Porti EMG system (TMS-international, Enschede, The Netherlands), with input impedance adapted to fine wire (CMRR > 90 dB, band-pass filtering between 20 Hz and 1 kHz, and 22 bits AD conversion after 20 × amplification).

Kinematics

Four cluster markers were attached halfway along the upper and lower legs. Each cluster marker included three infrared emitting diodes for movement registration with a 2 × 3 camera system (OPTOTRAK 3020, Northern Digital, Waterloo, Ontario, Canada), connected via a synchronization cable to the Porti5 EMG system. For kinematic data (cf. Fig.1 bottom row), the sampling frequency was 50 samples/s.

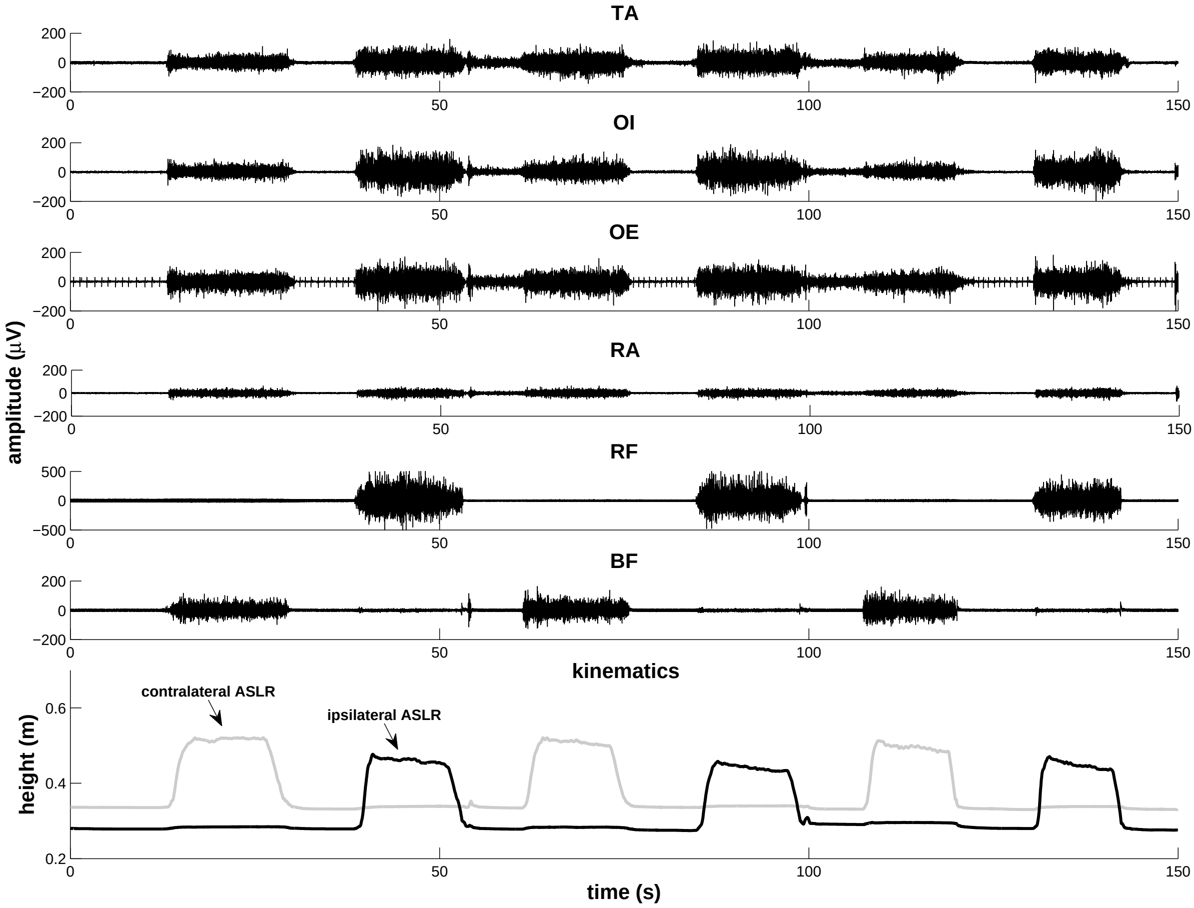

Fig. 1 Raw electromyograms, with signals (μV) over time (s), in one subject during three consecutive repetitions of contralateral and ipsilateral ASLR. Note the scale difference between RF and the other muscles. The bottom panel gives the kinematical pattern, with the height (m) of the leg raise over time. Each ASLR lasted around 10 seconds, and subjects rested about 10 seconds between repetitions. Baseline values of contralateral and ipsilateral ASLR are arbitrary. For the abbreviations of the muscle names, cf. Table 1.

Conditions

Ipsilateral (right) and contralateral (left) ASLR were performed in supine position with the legs straight and the feet in dorsiflexion, 20 cm apart (Mens et al., 2001). Subjects were instructed to raise each leg three times until the foot reached 20 cm above the table, without bending the knees, and keeping the leg elevated for around 10 seconds ("Normal"). After every ASLR, subjects were asked to relax for approximately 10 seconds. The whole procedure was repeated with a weight added just above the ankle ("Weight"), so that the static moment of the leg with respect to the hip was increased by 50%. To calculate (Zatsiorsky, 2002, p. 605) the required amount of weight, manually measured lower extremity anthropometry was taken into account. Finally, the ASLR was repeated with a non-elastic pelvic belt (3221/3300, Rafys, Hengelo, The Netherlands) around the pelvis ("Belt"). The pelvic belt was placed just below the spina iliaca anterior superior ("high position", Damen et al., 2002; Mens et al., 2006b), with a tension of 50 N (Vleeming et al., 1992; Mens et al., 1999), fine-tuned with an inbuilt gauge.

Data analysis

Data were analyzed with MATLAB 7.4 (The Mathworks, Nattick, MA, USA). Kinematic data were filtered with a 4th order bi-directional low pass Butterworth filter with a cutoff frequency of 5 Hz. Onset and peak of leg raise (first point with zero velocity before/after a peak in velocity) and leg raise velocity (height of peak position divided by the time to reach peak position) were derived. From the markers in the relevant clusters, mean heights of the upper and lower legs were calculated over the three repetitions per subject per condition.

TA EMG was not usable in four subjects, leaving twelve valid datasets. EMG data were high-pass filtered at 250 Hz (1st order Butterworth) to remove ECG contamination and to obtain less variable estimates of EMG amplitude (Potvin and Brown 2004; Staudenmann et al. 2007), then full-wave rectified, and low-pass filtered at 5 Hz (2nd order Butterworth). The median amplitude during ASLR plateau (5 through 10 s after movement onset) was calculated.

To quantify the asymmetry of activity of TA, OI, and OE, an Asymmetry Index (ASI) was calculated from EMG amplitudes during ipsi- and contralateral ASLR as: (ipsilateral – contralateral) / (ipsilateral + contralateral) × 100%. Positive values indicate more ipsilateral than contralateral activity, negative values more contralateral activity, and the index expresses the difference between the two as percentage of their sum.

Statistical analysis

SPSS 16 was used throughout, with $p < 0.05$ as threshold for significance. First, outliers were identified from box plots of muscle amplitude and of the Asymmetry

Index (Figs. 2 and 3), and removed. Then, Generalized Estimation Equations (GEEs) were used, i.e., repeated measures regression analyses that allow for missing values. GEEs assessed the effects of “Factors” on the dependent variables, i.e., EMG amplitude or the Asymmetry Index. As non-normalized EMG amplitudes of different muscles cannot be compared, effects on EMG amplitude were assessed for each muscle separately. This was not required for the Asymmetry Index.

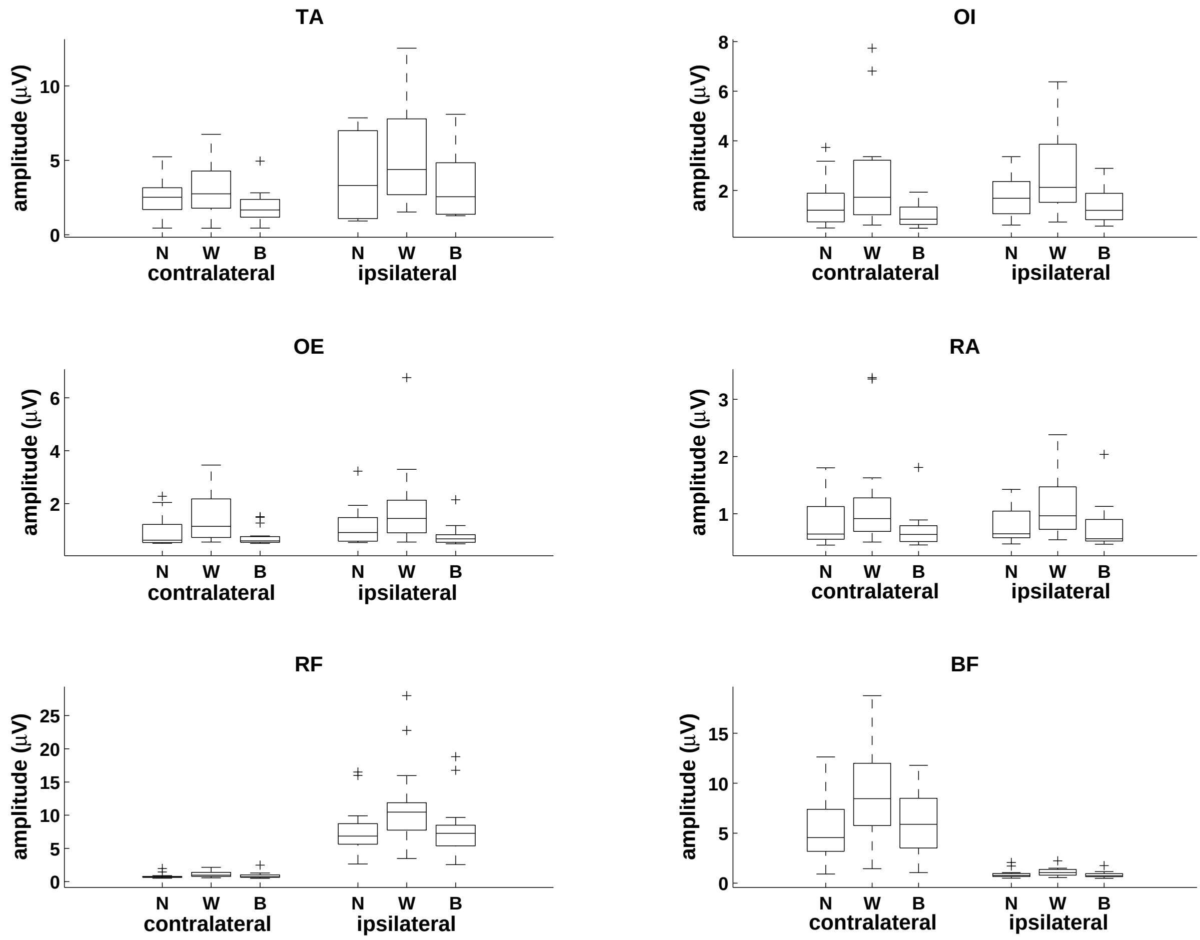

Fig. 2 Box plots of median muscle activity (μV) during ipsilateral and contralateral ALSR in three conditions (N = normal, W = with weight added, B = with a pelvic belt). Each box runs from the 25 to the 75 percentile; the transverse line inside the box indicates the median, “+” represents outliers, and the error bars represent the range, excluding the outliers. Note the scale differences. For the abbreviations of the muscle names, cf. Table 1.

Factors included in the analysis were Side (Ipsilateral, or Contralateral), Condition (Normal, Weight, or Belt), and Muscle. GEEs perform nested calculations, first of Model Effects (cf. “Model” in Tables 1 and 2), which assess the general impact of each Factor, then Parameter Estimates (cf. “Parameters” in Tables 1 and 2), which compare each specific value of that Factor to a reference. The

Contralateral Side and the Normal Condition were used as references. The Asymmetry Index was only tested for muscles that had a significant effect of Side on muscle activity, and the muscle with the smallest median asymmetry in a box plot (cf. Fig. 3) was used as reference.

Table 1 *P*-values (bold when significant) and significant regression coefficients (*B*) from (top four muscles) GEEs on abdominal muscle activity during ipsilateral and contralateral ASLR, with Side and Condition as Factors, including their interaction, and also from (bottom two muscles) GEEs on RF activity during ipsilateral, and of BF activity during contralateral ASLR, with Condition as factor. Note that GEEs calculate regression equations, and, for instance, the second line reads as: TA activity (μV) = 2.56 - 0.86 (in the condition with the belt) + 1.24 (in the condition with weight added during the ipsilateral ASLR), and some effects that were not significant in these parameter estimates.

Muscle (μV)	Intercept *p*	Intercept *B*	Side[a] *p*	Side[a] *B*	Condition[b] *p*	Condition[b] *B*	Interaction[c] *p*	Interaction[c] *B*
Abdominal muscles:								
TA								
Model	**0.00**		**0.00**		**0.01**		**0.01**	
Parameters	**0.00**	2.56	0.07		W: 0.11		Ipsi×W: **0.03**	1.24
					B: **0.01**	-0.86	Ipsi×B: 0.12	
OI								
Model	**0.00**		**0.02**		**0.00**		n.s.	
Parameters	**0.00**	1.29	**0.02**	0.53	W: **0.00**	0.74		
					B: **0.00**	-0.38		
OE								
Model	**0.00**		0.42		**0.00**		n.s.	
Parameters	**0.00**	0.98			W: **0.00**	0.57		
					B: **0.00**	-0.23		
RA								
Model	**0.00**		0.28		**0.00**		n.s.	
Parameters	**0.00**	0.80			W: **0.00**	0.23		
					B: **0.01**	-0.15		
Leg muscles:								
RF (tested for right ASLR only)								
Model	**0.00**				**0.00**			
Parameters	**0.00**	6.40			W: **0.00**	2.85		
					B: 0.52			
BF (tested for left ASLR only)								
Model	**0.00**				**0.00**			
Parameters	**0.00**	5.40			W: **0.00**	3.59		
					B: **0.00**	0.83		

[a] The *B*-value is for ipsilateral compared to contralateral activity

b	Comparing Weight (W) or Belt (B) with the Normal condition

c	*B*-values are for ipsilateral (Ipsi) compared to contralateral activity in the Weight (W) or the Belt (B) condition, both compared to Normal

TA:	m. transversus abdominis

OI:	m. obliquus internus abdominis

OE:	m. obliquus externus abdominis

RA:	m. rectus abdominis

RF:	m. rectus femoris

BF:	m. biceps femoris

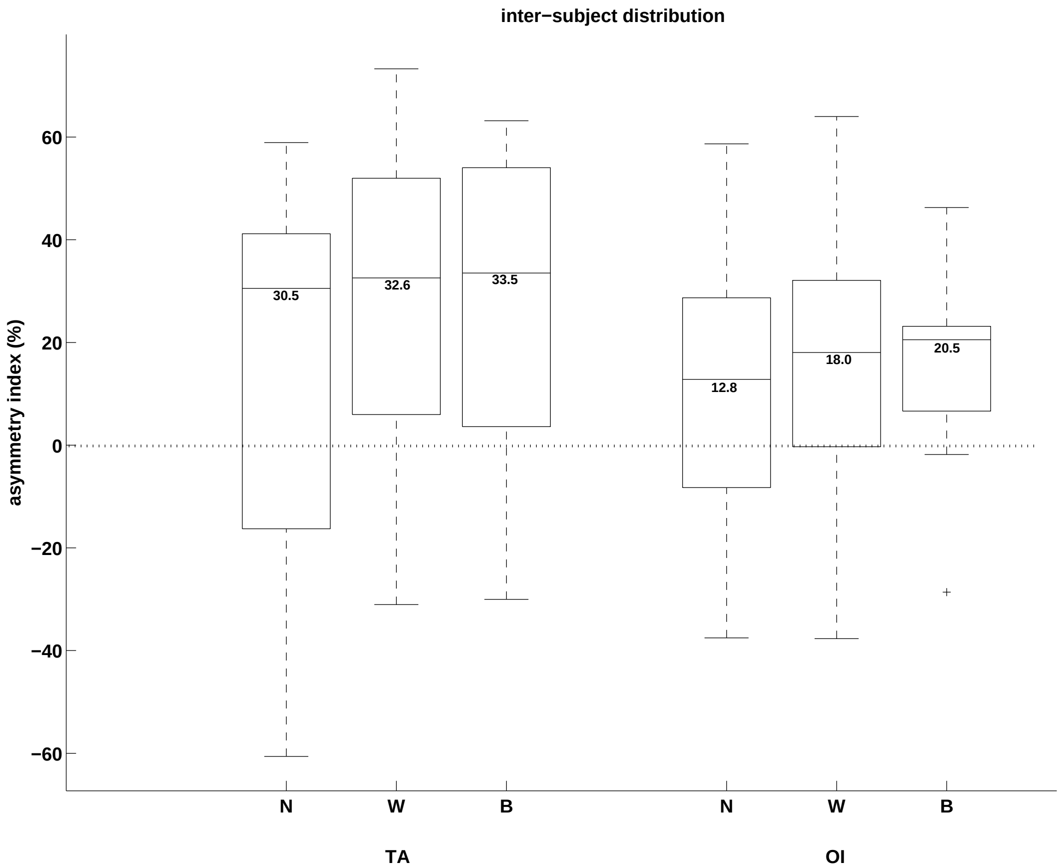

Fig. 3 Box plots of the Asymmetry Index (%) for TA and OI in all three conditions (cf. Fig. 2 & Table 1).

Results

Heights reached by the upper or lower leg were not significantly affected by Side or Condition, nor was there a significant interaction. The maximum velocity of leg raise was affected by Condition (p < 0.001); movement was faster with the application of the belt (0.25 m/s), and slower with weight (0.22 m/s) than in the normal condition (0.23 m/s). There was no significant effect of, or interaction with, Side.

Table 2 *P*-values (bold when significant) and significant regression coefficients (*B*) from GEEs on the Asymmetry Index of the TA compared to OI, with Condition and Muscle as Factors, and including their interaction (cf. Table 1).

Asymmetry Index	Intercept		Condition			Muscle		Interaction		
(%)	*p*	*B*		*p*	*B*	*p*	*B*		*p*	*B*
TA vs OI Model	**0.00**			**0.02**		0.26			**0.00**	
Parameter	**0.00**	10.73	W:	0.89		0.82		W×TA:	**0.00**	15.99
			B:	0.30				B×TA:	0.26	

Muscle activity

Fig. 1 provides a typical example of EMG activity. All abdominal muscles were bilaterally active, but activity was ipsilateral in RF and contralateral in BF. Thus, we analyzed the effects of Side, Condition, and their interaction, for all abdominal muscles, whereas we only analyzed the effects of Condition on RF during ipsilateral, and on BF during contralateral ASLR.

Tests of model effects on the abdominal muscles (Table 1, cf. Fig. 2) revealed one significant interaction, in TA, with parameter estimates indicating more ipsilateral activity when weight was added (B = 1.24, p = 0.03). There was an effect of Side for OI, with more ipsilateral than contralateral activity (B = 0.53, p = 0.02), a tendency for the same difference in TA (B = 1.26, p = 0.07), more activity in conditions with weight (but not in TA, where the interaction was significant), and less with the pelvic belt.

Ipsilateral RF activity (Table 1) was larger with weight (B = 2.85, p = 0.00), as was BF activity (B = 3.59, p = 0.00), which was, perhaps surprisingly, also larger with the pelvic belt (B = 0.83, p = 0.00).

Symmetry/asymmetry of TA and OI activity

Table 1 shows a significant effect of side on TA and OI activity, but not on OE and RA. Hence, we further assessed asymmetry of TA and OI activity only. Box plots (Fig. 3) suggested that most, but not all, subjects had more ipsilateral than contralateral activity, and median asymmetry appeared larger in conditions with weight added and with the pelvic belt, than in the normal condition. These results are in agreement with our expectations.

The median Asymmetry Index varied between 12.8% and 20.5% in OI, and between 30.5% and 33.5% in TA (Fig. 3). Thus, we used OI as reference in the GEE (Table 2). In the test of model effects, the effect of condition was significant, but in the parameter estimates, neither Weight nor Belt was significantly different from Normal. The difference between the muscles was not significant, but there was a significant interaction, TA activity with weight added was significantly more asymmetrical than OI (*B* = 15.99, *p* = 0.00).

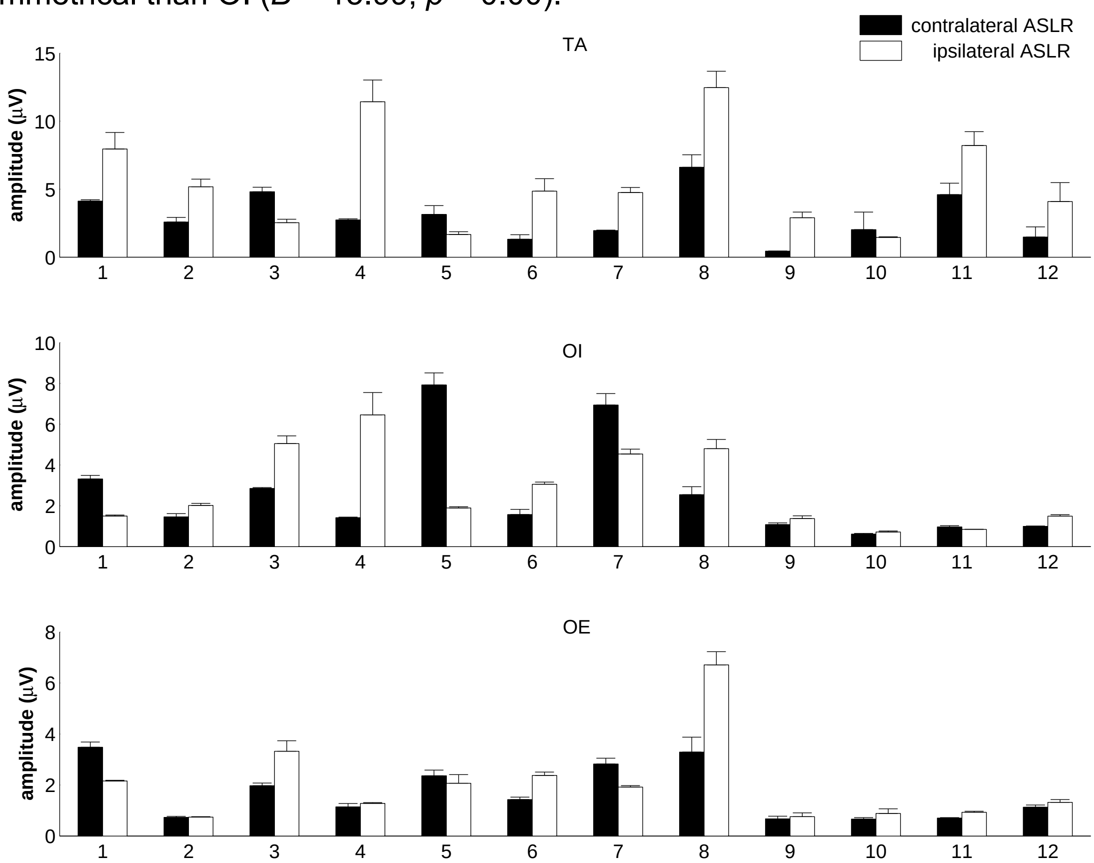

Fig. 4 TA, OI, and OE activity during contralateral (black) and ipsilateral (white) ASLR, for each individual subject averaged over the three repetitions in the condition with weight. Error bars represent standard errors. In cases where the error bar is not visible, the standard error approached zero.

Inter-individual differences

The box plots (Figs. 2 and 3) suggested substantial differences between subjects, especially in conditions with weight (Fig. 2). We depicted activity of the oblique and transverse abdominal muscles for all 12 subjects with a complete data set (Fig. 4), averaged over the three repetitions of contralateral and of ipsilateral ASLR, all with weight added. Average TA activity during ipsilateral ASLR was larger in 9 subjects, OI in 8, and OE in 9. Of these, 6 subjects had more activity during ipsilateral ASLR in all three muscles. On the other hand, one subject had more activity during contralateral ASLR in all three muscles.

Discussion

Toward a 3-D understanding of the ASLR

On initial impression, the ASLR is a unilateral task, but only RF EMG agreed with this simplified understanding, with activity only during ipsilateral ASLR, and more so when weight was added. On the other hand, BF behaved in clear opposition to any simplified view of the ASLR, with activity only on the side contralateral to the lifted leg, more so when weight was added, and, perhaps surprisingly, also with the pelvic belt.

During the ASLR, hip flexors may not only raise the leg, but also pull the ipsilateral ilium forward (Mens et al., 1999; Hu et al., 2010; cf., e.g., Vleeming et al., 1992, 1996, 2008; Hungerford et al., 2004). Earlier, we reported (Hu et al., 2010) that this forward rotation is prevented by contralateral BF activity, which pulls the contralateral ilium backward. Such BF activity is only useful if the two sides of the pelvis are pressed together, i.e., “force closure”, so that the force of the biceps can be transferred towards ipsilateral, with the pelvis moving in the sagittal plane as one unit (Vleeming et al., 1990a & b; Snijders et al., 1993a & b). Significant effects of side were found in TA and OI, but we conclude that the activity of all lateral abdominal muscles contained a symmetric component (“+ = +” in Table 3), as evident in the decrease of TA, OI, and OE activity when a pelvic belt was used.

Table 3 Plausible roles of the abdominal muscles during the Active Straight Leg Raise.

task component	RA			OE			OI			TA		
	I		C	I		C	I		C	I		C
force closure	-		-	+	=	+	+	=	+	+	=	+
posterior rotation of the pelvis	+	≥	+	+	≥	+	+	≥	+	-		-
ipsilateral rotation of the trunk	-		-	-		-	(+)		-	+		-

I: Ipsilateral C: Contralateral

- no role (+) possibly a role + clearly a role

I = C symmetrical task component

I ≥ C ipsilateral activity similar to, or larger than contralateral activity

RA was also bilaterally active (although with very low amplitude), more so with weight added, and less with the pelvic belt. When force closure is exerted, bilateral RA, OE, and OI activity will rotate the pelvis posteriorly, that is, thus contribute to counteract the forward rotation of the ipsilateral ilium. With a pelvic belt, the abdominal muscles are less active, implying less backward rotation of the pelvis, which may explain why BF is more active in conditions with a belt.

Note that abdominal muscle activity to rotate the pelvis backward may be symmetric, but it is the ipsilateral ilium that is being pulled forward, and, with sub-maximal force closure, it is quite possible that backward rotation of the pelvis involves more ipsilateral than contralateral activity ("+ ≥ +" in Table 3; cf. Beales et al., 2009a). Still, it is unclear why RA was less active in conditions with a pelvic belt.

Contralateral BF activity presses the contralateral heel against the bench, as reported previously (Beales et al., 2009a, b, 2010a), with more pressure when weight was added to the lifted leg (Beales et al., 2010b). Pressing down the contralateral heel will cause the contralateral side of the pelvis to move upwards, that is, ipsilateral rotation of the pelvis, as often observed clinically, and reported by Liebenson et al. (2009). Contralateral pelvis rotators (= ipsilateral trunk rotators), i.e., TA and OI (Urquhart and Hodges, 2005; Hu et al., 2010), will counter ipsilateral pelvis rotation. Beales et al. (2010b) did not measure TA, but reported increased ipsilateral OI activity when weight was added, whereas in the present study, TA was more asymmetric than OI (Figs. 2 & 3), particularly when weight was added (Table 2). We conclude that counterrotation is an important role of these two muscles in the

ASLR, with, in the present study, more involvement of TA than of OI (Table 3), probably because OI would exert a greater unwanted trunk lateroflexion moment, while TA fibers are mainly horizontal.

The above biomechanical understanding implies that some lateroflexion moment towards ipsilateral remained present. Lateroflexion has been mentioned as a theoretical possibility (Mens et al., 1999), but we neither measured nor observed it. Resistance between the thorax and the bench may have been sufficient to counter trunk lateroflexion.

The notion of "symmetry"

Authors tend to report "symmetry" when statistical analysis does not reveal a significant effect of side (e.g., Danneels et al., 2001; Beales et al., 2010b). Strictly speaking, this is not accurate, because one cannot prove exact symmetry. More important, this tendency distracts from the fact that muscles engage in multitasking (Saunders et al., 2004; Hu et al., submitted), with some task components being symmetrical, and others asymmetrical (Hodges, 2008).

"Symmetry" is a mathematical concept (De Sautoy, 2008), and symmetry may be a property of tasks, as understood biomechanically, not an empirical property of muscle activity or shape. Theoretically, force closure implies symmetric TA, OI, and OE activity. On the other hand, the lack of a statistical effect of side on OE (Table 1) does not prove that OE was only engaged in force closure, as it may have played a role in backward rotation of the pelvis. All four abdominal muscles (Table 3) have different symmetric and asymmetric task components. RA, for instance, had no intrinsically symmetric task component but had empirically no effect of side, and TA was expected to have a clear symmetric task component, but was the most asymmetrically active muscle.

Variability

There were striking inter-individual differences in the predominance of ipsilateral or contralateral activity of TA, OI, and OE. Fig. 4 appears to imply that most subjects had significant asymmetry in TA, OI, and OE activity, notwithstanding the presence of a symmetric task component. Motor control is intrinsically noisy (Van Gemmert and Van Galen, 1997), with limited precision (Georgeopoulous et al., 1993), and when subjects perform the same task repeatedly, there are large variations of the force produced (Van Dieën et al., 2001). More important, the redundancy of the motor system invites subjects to flexibly deal with the degrees of freedom, and motor strategies tend to be stochastically selected (Bernstein, 1967; cf. Latash et al., 2002; Meijer and Bruijn, 2007). Beales et al. (2010b) reported more ipsilateral OI activity in the ASLR when weight was added, but this was not significant in the present study, which may have been due to variations in the selection of strategies, especially with respect to the transverse and frontal planes. Further ASLR research aiming to understand the ASLR should focus on 3D kinematics and kinetics, including the role of pelvis rotation, and subjects' strategies to avoid trunk lateroflexion.

Limitations

The most important limitation of the present study was that pelvis and trunk kinematics was not measured. Further, muscles were only studied on the right side, although right and left ASLR were performed. Four sets of TA data could not be used, and outliers were removed before statistical testing. The use of surface EMG for OI and OE in the present study may have affected results. Crosstalk between the OI and OE signals, and between TA and OI, cannot be excluded. On the other hand, fine-wire EMG of TA would only reflect the activity of the mid region of that muscle, whereas different functional roles of different parts of TA have been described (Urquhart and Hodges, 2005). Finally, we only measured women, and generalization of our results to the male population may not be straightforward.

Conclusion

The abdominal muscles were bilaterally active in the Active Straight Leg Raise (ASLR). Statistically, there was a significant difference between TA and OI EMG during right compared to left ASLR, with most asymmetric activity in TA when weight was added above the ankle. The four abdominal muscles were differently involved in different tasks, some tasks being intrinsically symmetrical, some unilateral only. During the normal ASLR, the forward pull on the pelvis by hip flexors must be countered by contralateral hip extension and/or activity of the abdominal muscles;

symmetrical force closure must ensure that the pelvis moves as one unit in the sagittal plane, and a balance is needed between ipsi- and contralateral rotation of the pelvis. “Symmetry”, thus, is a property of certain tasks, and not an empirical property. Variability was large, and to better understand the ASLR, future research should include the 3D kinematics and kinetics of the pelvis and the trunk.

Chapter 6

Epilogue

Motor control in the lumbopelvic region

The central nervous system controls the musculoskeletal system to coordinate its multiple degrees of freedom, organizing movements into skilled actions, and taking into account the interactions between the individual, the task and the environment. The present thesis focused on motor control in the lumbopelvic region, which, although it allows limited movement in most tasks, is a dynamic system, supporting the weight of the upper body, and transferring loads between the trunk and the lower limbs. With the clinic in mind, especially the puzzling problem of Pelvic Girdle Pain (PGP), we limited ourselves to healthy subjects, reasoning that the understanding of pathophysiology requires the understanding of normal physiology first.

Stable behaviour, as accomplished by the intricate interplay between external forces and lumbopelvic muscle activity within a complex three-dimensional geometry, is critical for the lumbar spine and the pelvis to bear loads, allow movement, and at the same time avoid injury and pain. The lumbar spine and pelvis are inherently unstable: The osteo-ligamentous lumbar spine in vivo buckles under a compressive load of only 90N (Crisco et al., 1992), and the sacroiliac joints with their relatively flat surfaces are vulnerable to shear forces (Snijders et al., 1993a & b). Note that these joints cannot be stabilized through passive, ligamentous forces, as continuous loading of ligaments would lead to creep (Richardson et al., 2002; Pel et al., 2008).

We expected that stability would result from highly coordinated muscle activation patterns involving many muscles. Still, stability is only one of several tasks that the control system has to face simultaneously. By now, we understand that lumbopelvic stability requires bilateral, symmetrical muscle activity, on top of which other tasks may require unilateral activity, with a net result that is asymmetrical. Thus, EMG demonstrates various patterns of laterality during the ASLR and in gait. Therefore, one should take the three dimensional kinematics and kinetics of multiple body segments into account to gain a more in-depth understanding of motor control in the lumboeplvic region.

Bilateral muscle activity may contribute to lumbopelvic stabilization

Muscles in the lumbopelvic region clearly showed bilateral activity, which implied (but did not prove) their contribution to lumbopelvic stabilization. Still, the fact that less activity was observed of the oblique and transverse abdominal muscles during an ASLR with a pelvic belt (Chapter 2), may be taken to present proof of a stabilizing role of these muscles, as suggested in Snijders' theory of force closure (Vleeming et al., 1990a & b; Snijders at al., 1993a & b).

In the literature, the psoas is regarded as (mainly) a hip flexor, but in our own study, it was active during ipsilateral as well as contralateral ASLR, which suggests a role in stabilization. When weight was added above the ankle, there was more psoas activity, which, however, remained as bilateral as in a normal condition (Chapter 3). Maybe this persistence of the basic pattern under different conditions can be taken as proof that the psoas acts as a stabilizer during the ASLR, providing frontal plane stability to the lumbar spine. Such psoas activity will add to compressive forces, which stiffen spinal joints (Santaguida and McGill, 1995).

Moreover, the rectus abdominis and erector spinae were bilaterally active during ASLR, without a significant effect of side (right or left ASLR), which suggests (but does not prove) that they were involved in sagittal plane stabilization of the pelvis and lumbar spine.

The importance of multitasking

The abdominal muscles are involved in several tasks, such as stabilizing the spine and pelvis, breathing, control of trunk movement in three dimensions, preparing for potentially destabilizing events such as heel contact, or laughing, coughing, emptying the bladder and bowel movements. Stabilizing muscle activity usually implies a symmetric task component, but actual muscle activity may well be asymmetrical, since other tasks may require unilateral activity (Chapters 4 and 5). So, in many instances, we will find asymmetric components on top of symmetric components. This was found, for instance, for contralateral pelvis rotation by TA (asymmetric) on top of a role in force closure of TA (symmetric) in the ASLR (Chapter 5). When lifting one leg in ASLR, subjects exert contralateral downward pressure (Beales et al., 2009a) with their biceps femoris, leading to a moment on the pelvis that would rotate it ipsilaterally, but is resisted by TA activity and, according to data presented in chapter 5, to a lesser extent, OI activity.

In treadmill walking, the lateral abdominal muscles showed an intricate pattern of unilateral as well as bilateral activity (Figure 1), which could only be unravelled by separate analyses of the different task components of the main tasks these muscles were engaged in. In our analyses of walking, and of the ASLR (with a pelvic belt or with weight added), it turned out to be very useful to distinguish movement and stabilization in the sagittal, the transverse, and the frontal plane. In the ASLR, for instance, contralateral biceps femoris activity and force closure are needed for sagittal plane stabilization of the pelvis, while ipsilateral rotation of the pelvis then has to be balanced by contralateral rotators (transverse plane), which, however, may cause lateroflexion of the trunk (frontal plane).

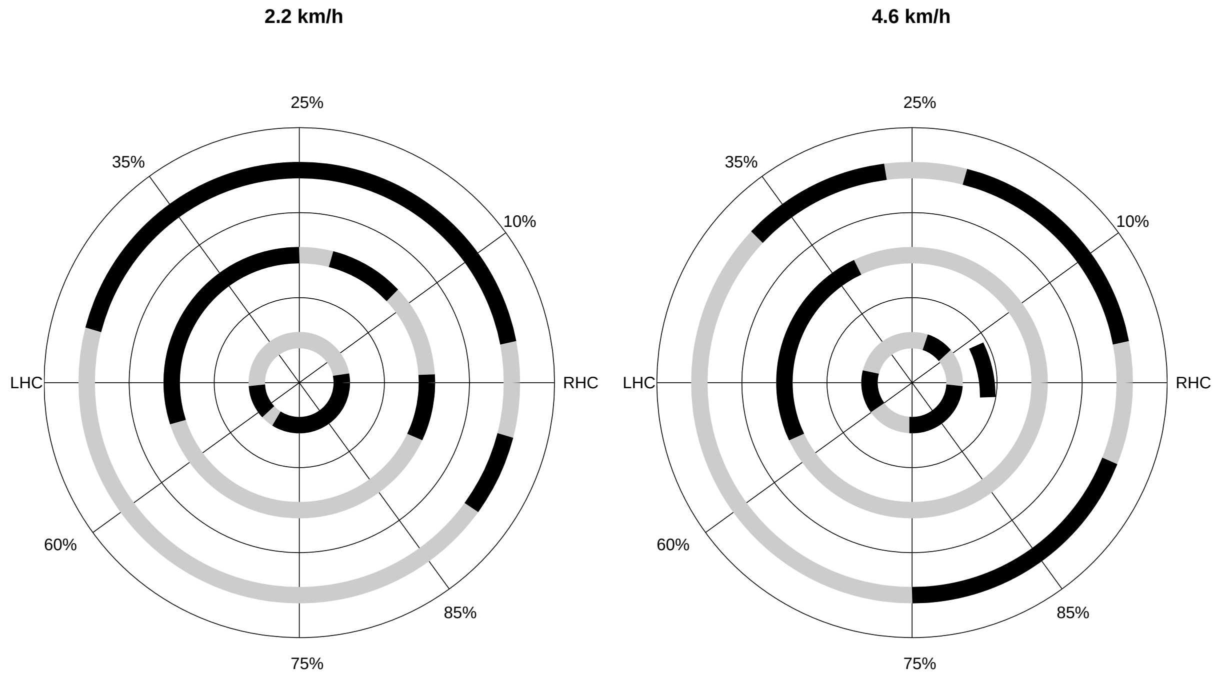

Fig. 1 Major muscle activity (black) of TA (inner ring), OI (in-between) and OE (outer ring) during walking at 2.2 km/h (left), and 4.6 km/h (right).

No single muscle is limited to a single task

The studies in the present thesis suggested that no muscle is limited to a single task (see also Hodges, 2008). Transversus abdominis has been considered as an important muscle for the control of lumbopelvic stability, with a role that “differs from the other abdominal muscles” (Hodges, 1999, p. 74). At first sight, this may appear to disagree with Cholewicki and VanVliet’s conclusion that “A single muscle cannot be identified as the most important for the stability of the lumbar spine. Instead, the relative contribution of each muscle to spine stability depended on trunk loading direction and magnitude.” (Cholewicki and VanVliet, 2002, p. 99). Note, however, that there are two sides to this argument. First, a claim that transversus abdominis is always used for stabilization purposes only was never made (Hodges, 2008 vs. Allison et al., 2008), and would be wrong, as our studies clearly illustrate. Second, there is no doubt that some muscles have at least one mechanical effect that is unique to them, and the claim that transversus abdominis may increase stiffness between lumbar vertebrae in a way that differs from other muscles (Hodges, 1999) is theoretically conceivable, and requires further investigation. The studies of the present thesis, with its focus on pelvic stability, did not assess such a possible function of deep trunk muscles.

Muscle coordination to deal with conflicting constraints

It will be rare for any individual muscle to have the exact mechanical effect required for the task at hand. Thus, as a rule, muscle activity will have unwanted side-effects. The studies of the present thesis presented many examples of the way the control system deals with such side effects. Raising a leg from a supine position, for instance, requires hip flexor activity, which will pull the ilium forwards. Given force closure, the contralateral biceps femoris may then be used to counter this forward pulling. Such activity of the biceps femoris may lead to ipsilateral rotation of the pelvis, which is then countered by activation of contralateral pelvis rotators. So, in general the control system will not select single muscles to perform a particular task, but two, three, or even more.

The variability of motor performance

Task constraints determine which variables are the most important ones in performing a task. The ASLR, for instance, requires ipsilateral hip flexion, which is essential to this task, and in general, no ipsilateral hip extensors will be used during the ASLR, unless stability (Cholewicki et al., 1997) or precision (Selen et al., 2006) requirements are best met by co-contraction. Other degrees of freedom may be less crucial. Trunk orientation, for instance, is not precisely prescribed in the ASLR. Thus, subjects may select different strategies, in particular with regard to the variables that are not essential. In the ASLR (Chapter 5), we found clear evidence of such variability. Clinically, the ASLR is often interpreted in terms of muscle function. To allow for such an interpretation, more complete kinematic and kinetic control over or analysis of the ASLR is required.

Future directions of research into pelvic girdle pain (PGP)

The present study limited itself to healthy subjects, in the hope that better insight into normal physiology would lead to a better understanding of possible mechanisms underlying PGP. The overriding conclusion from all studies of the present thesis is that force closure is of paramount importance to pelvic stability, at least in any task that requires hip flexion, such as the ASLR or walking. Thus, the hypothesis presents itself that PGP may arise from failing force closure.

If force closure does *not* work during hip flexion, the ilium will be pulled forwards. Note that psoas activity would have an opposite effect on the ilium, but it was shown not to be active as a hip flexor during the ASLR (Chapter 3). So, the hip flexors that are used during the ASLR will pull the ilium forwards. Such forward rotation of the ilium will lead to stretching of the long dorsal sacroiliac ligaments, which connect the ilium with the sacrum. The long dorsal ligaments have been implicated in PGP, and are often painful on palpation in patients (Vleeming et al., 2002). Moreover, forward movement of the ilium implies backward movement of the sacrum, i.e., counternutation, and with the sacrum in counternutation, the sacroiliac joint is in its most unstable position (Vleeming et al., 2008). Theoretically, hip flexion could then lead to sagittal plane sacroiliac torsion and subsequent micro-injury, akin to Panjabi's "ligament subfailure injuries" in low back pain (Panjabi, 2006). Unfortunately, such injuries do not reveal themselves when imaging techniques are used, such as MRI, at least not with current methodology. Still, there is some evidence that local sacroiliac injury plays a role in at least some forms of PGP. Clinical experience clearly suggests that sacroiliac injection of pain killers is helpful in some patients (Schwarzer et al., 1995).

Since the pelvis is a unit, sacroiliac instability will imply instability of the symphysis. Indeed, vertical movements of the pubic bones with respect to each other, suggesting sagittal plane rotation in the SI joints, have been observed (Mens et al., 1999). Interestingly, the symphysis area appears to be the locus of a second kind of pathology, which was discovered in recent years to occur in athetes with chronic groin pain (Verrall et al., 2001). The leading idea is that athletes may overload their adductors, leading to painful origin tendons of the adductors, with, eventually, a bone marrow inflammation of a pubic bone, i.e., *osteitis pubis*. Such osteitis pubis does show itself on MRI, and we observed it in a 30-years-old woman not long after the delivery of her first child (Figure 2).

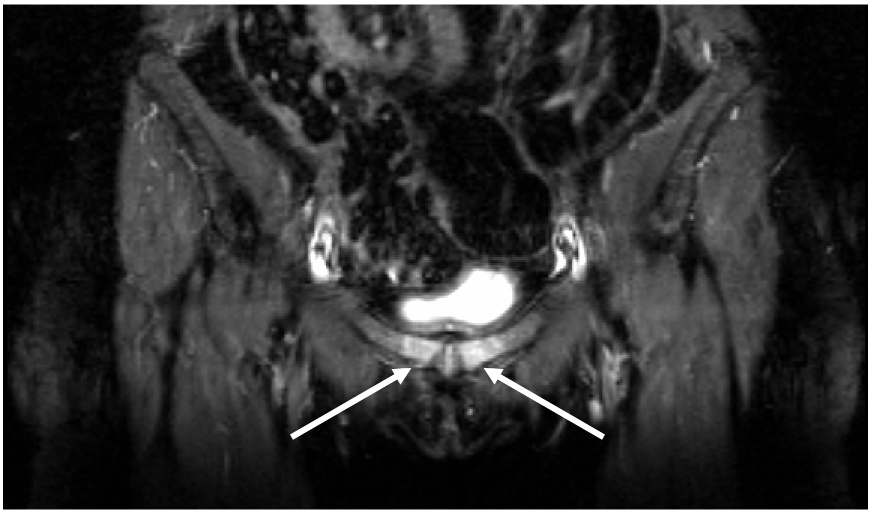

Fig. 2 Post-partum osteitis pubis (T2/STIR MRI).

From the viewpoint of the present thesis, the above ideas on possible consequences of failing force closure during hip flexion are speculative. But, they are in agreement with the evidence. Hence, two questions remain. First, if force closure is indeed lacking in PGP patients. Second, if that would explain the effects of a pelvic belt. As to the first question, a recent ultrasound study revealed less increase of the thickness of TA and OI in healthy athletes in anticipation of pain by electrical stimulation in the groin area (Jansen et al., 2010a). Somewhat surprisingly, this did not happen during actual pain. Moreover, OE showed an increase of thickness during anticipation of pain. Less TA and OI activity when anticipating pain may suggest some form of reflex inhibition (Hurley and Newham 1993). Accordingly, a failure of force closure may occur in sacroiliitis or symphysitis, that is, in situations where pelvic compression is likely to hurt. In athletes with chronic groin pain, decreased resting thickness of TA and OI was reported (Jansen et al. 2010b). Clearly, the default hypothesis would now be that force closure muscles are not recruited as long as compression of the pelvis would be painful, the muscles then atrophy, and even after the inflammation has ended, not enough force closure can be produced. Again, this is speculative, and again, this is in agreement with evidence. Still, it remains unclear why TA and OI would be affected and OE not. It has been argued that ultrasound is insufficiently sensitive to assess thickness changes of OE (Hodges et al. 2003). Moreover, TA and OI belong to the “deep” muscles that are later and less active in anticipating perturbations in low back pain patient (e.g., Hodges and Richardson 1999), and multifidus atrophy has been reported (Hides et al. 1994), but the underlying mechanisms remain poorly understood.

There is little doubt that PGP leads to a vicious cycle, with initial damage causing more damage, etc. Thus, one would expect a pelvic belt to only relieve pain that is the direct result of failing force closure. In a study of athletes with chronic groin pain, 20% of the subjects had less pain during an ASLR when a pelvic belt was applied, but the difference between responders and non-responders was not related to thickness of TA and OI. (Jansen et al. 2010c). Nor, we argue, should such a relationship be expected, because in a vicious cycle it is the present cause of pain that counts. Clearly, further research will be needed to establish when a pelvic belt can be, and is, helpful. Moreover, one may expect a pelvic belt to increase pain in acute sacroiliitis, because it would lead to compression. For acute symphysitis, this is less clear, since one can imagine that compression at the level of, say, the iliac crests would decompress the symphysis, while compression at a lower level would compress it. For overuse injury of the long dorsal ligament with insufficient force closure, a belt would likely be effective as it would compensate for dysfunction of the abdominal muscles, stabilize the SI joints, and reduce loading of this ligament. A more detailed biomechanical understanding of load transfer within the pelvis will be needed. Anyhow, for me, this is enough speculation. Time to go to work.

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Summary/Samenvatting

Motor Control and Lumbopelvic Stability in Young Healthy Women

Many people, especially women during or after pregnancy, suffer from Pelvic Girdle Pain (PGP), which has remained a puzzling problem for a long time. With clinical relevance in mind, the present thesis limited itself to research on young healthy women, reasoning that the understanding of pathophysiology requires the understanding of normal physiology first. Chapter 1 presents a general introduction, focusing on the relationship between motor control and lumbopelvic stability.

Chapter 2 takes its starting point in Snijders' assertion that the sacroiliac joints are intrinsically unstable, and that concerted activity of the lateral abdominal muscles is needed to press the ilia against the sacrum, i.e., *force closure*. This mechanism may be disturbed in PGP. Electromyographic (EMG) activities during the Active Straight Leg Raise (ASLR), and during treadmill walking, were studied without or with a pelvic pelt, which is supposed to substitute part of force closure. All muscles measured were active during the ASLR. In both tasks, the lateral abdominal muscles were less active in conditions with the belt, which result confirms Snijders' theory of force closure. Increased activity of contralateral hip extensors was found, i.e., the m. biceps femoris during the ASLR, and the m gluteus maximus in treadmill walking. It is important to note that hip flexors exert a forward rotating torque on the ilium. As long as the lateral abdominal muscles press the ilia against the sacrum (Snijders' "force closure"), the pelvis moves as one unit in the sagittal plane, and contralateral hip extensor activity will contribute to preventing forward rotation of the ipsilateral ilium.

Psoas function is a topic of considerable relevance in sports and clinical science, but the literature on psoas function is insufficiently consistent. Chapter 3 focuses on the question if the m. psoas is mainly a hip flexor, like the m. iliacus, or more involved in stabilization of the lumbar spine, which would require bilateral rather than unilateral activity. EMG activity was measured of right-sided psoas, iliacus, rectus femoris, and adductor longus, all during left or right ASLR. The four muscles measured started to contract before movement onset. The iliacus, rectus femoris, adductor longus, and psoas were active ipsilaterally, but psoas was also active contralaterally. There was no significant difference between the amplitudes or onset times of ipsilateral and contralateral psoas EMG activity. Nor was there a significant

interaction between Side (left or right ASLR) and Condition (without or with weight added above the ankle) for the psoas. Although ipsilateral psoas activity is consistent with the psoas being involved in hip flexion, contralateral activity is not. The most plausible explanation of the pattern found is that the psoas is recruited to stabilize the lumbar spine in the frontal plane.

Chapter 4 focuses on multitasking of the lateral abdominal muscles during walking, and on the question how the control system deals with conflicting constraints. Transversus abdominis (TA), obliquus internus (OI), and obliquus externus (OE) are involved in multiple functions: breathing, control of trunk orientation, and stabilization of the pelvis and spine. We studied EMG activity of these muscles, and 3-dimensional moments during treadmill walking at six different speeds (1.4-5.4 km/h) in healthy young women. PCA revealed that the time series of trunk moments were consistent across speeds and subjects, though somewhat less in the sagittal plane. All three muscles were active during ≥ 75% of the stride cycle, which is indicative of a stabilizing function. Clear phasic modulations were observed, with TA more active during ipsilateral, and OE during contralateral swing, while OI activity was largely symmetrical. Fourier analysis revealed four main frequencies in muscle activity: respiration, stride frequency, step frequency, and a triphasic pattern (probably a summation of stride and step frequency). With increasing speed, the absolute power of all frequencies remained constant or increased; the relative power of respiration and stride-related activities decreased, while that of step-related activity and of the triphasic pattern increased. Effects of speed were gradual, and EMG linear envelopes had considerable common variance (> 70%) across speeds within subjects, suggesting that the same functions were performed at all speeds. Further analyses in the time domain revealed both simultaneous and consecutive task execution. To deal with conflicting constraints, the activity of the three muscles was clearly coordinated, with co-contraction of antagonists offsetting unwanted mechanical side-effects of each individual muscle.

Chapter 5 presents a more detailed understanding of muscle activity during the ASLR. In the literature, it is often assumed that stabilizing activity reveals itself as bilateral symmetry. Detailed analysis of the ASLR suggests that this is incorrect, and that actual muscle activity results from the summation of symmetrical and asymmetrical components. In Chapter 2, contralateral activity of a hip extensor was found, i.e., the m. biceps femoris. Such activity will press the contralateral heel downwards, and the contralateral pelvis upwards, which implies ipsilateral rotation of the pelvis. This ipsilateral rotation of the pelvis was mainly prevented by ipsilateral activity of the transversus abdominis, which offers another, more complex, example of the control system dealing with conflicting constraints by co-contraction, with unwanted effects of one muscle being offset by the activity of another muscle.

Chapter 6 presents a general discussion of motor control during the ASLR and during walking, in terms of lumbopelvic stability. By now, we know that bilateral activity of the lateral abdominal muscles may contribute to lumbopelvic stabilization; multitasking is important; no single muscle is limited to a single task; muscle coordination deals with conflicting constrains; and motor performance is variable. The most important conclusion of this thesis is that further biomechanical research into pelvic girdle pain needs to include the three dimensional kinematics and kinetics of (most of) the body. In and of itself, this conclusion may not be surprising, but so far, there is hardly any literature that takes the 3 dimensions involved sufficiently into account.

Bewegingssturing en lumbopelvische stabiliteit in jonge gezonde vrouwen

Pijn in de bekkengordel (PGP) komt veel voor, vooral bij vrouwen tijdens of na de zwangerschap, en is lange tijd niet of niet voldoende begrepen. Juist om klinische toepassing mogelijk te maken, focust deze dissertatie op onderzoek bij gezonde jonge vrouwen, aangezien het begrijpen van de pathofysiologie het begrijpen van de normale fysiologie vooronderstelt. In Hoofdstuk 1 wordt een algemene inleiding aangeboden, gericht op de relatie tussen bewegingssturing en lumbopelvische stabiliteit.

In Hoofdstuk 2 wordt het uitgangspunt genomen in Snijders' claim dat de sacroiliacale gewrichten intrinsiek onstabiel zijn, en dat gezamenlijke activiteit van de laterale buikspieren nodig is om de ilia tegen het sacrum te drukken, de zogenaamde *krachtsluiting*. Dit mechanisme zou in PGP verstoord kunnen zijn. Electromyografische (EMG) activiteit werd geregistreerd tijdens het Actief Strekken van het Geheven Been (ASLR), en tijdens lopen op een lopende band. Dit werd gedaan zonder of met een bekkenband, die verondersteld wordt ook krachtsluiting te kunnen leveren. Alle gemeten spieren waren actief tijdens de ASLR. In beide taken waren de laterale buikspieren minder actief wanneer de bekkenband werd gedragen, hetgeen een bevestiging is van Snijders' theorie van de krachtsluiting. Er werd toegenomen activiteit gevonden van de contralaterale heupstrekkers, namelijk de m. biceps femoris tijdens de ASLR en de m. gluteus maximus tijdens het lopen. Het is belangrijk om te beseffen dat de heupbuigers het ilium naar voren roteren. Zolang de laterale buikspieren de ilia tegen het sacrum aandrukken (Snijders' krachtsluiting), beweegt de pelvis in het sagittale vlak als een eenheid, en draagt contralaterale activiteit van heupstrekkers bij aan het voorkómen van een voorwaartse rotatie van het ipsilaterale ilium.

Het begrijpen van psoasfunctie is een belangrijk onderwerp in sport- en klinische wetenschappen, maar de literatuur over psoasfunctie is onvoldoende consistent. Hoofdstuk 3 focust op de vraag of de m. psoas vooral een heupbuiger is, zoals de m. iliacus, of meer betrokken is bij het stabiliseren van de lumbale wervelkolom, hetgeen bilaterale en niet slechts unilaterale activiteit zou vereisen.

EMG-activiteit werd gemeten aan de rechter mm. psoas, iliacus, rectus femoris, en adductor longus, gedurende een ASLR met het linker of met het rechter been. Nog vóór het begin van de beweging werden deze spieren alle actief. De iliacus, rectus femoris, adductor longus, en psoas waren allemaal ipsilateraal actief, maar de psoas ook aan de contralaterale zijde. Er was geen significant verschil tussen de amplitudes van ipsilaterale en contralaterale EMG-activiteit van de psoas, geen verschil tussen de tijdstippen waarop de spier werd geactiveerd, en ook geen significante interactie tussen Been (rechter of linker ASLR) en Conditie (zonder of met gewicht toegevoegd boven de enkel). Ipsilaterale psoas-activiteit past bij een rol voor die spier in het buigen van de heup, maar contralaterale activiteit niet. De meest voor de hand liggende interpretatie is, dat de psoas geactiveerd wordt om de lumbale wervelkolom in het sagittale vlak te stabiliseren.

De focus van Hoofdstuk 4 is op de simultane betrokkenheid bij een aantal taken, *multitasking*, van de laterale buikspieren tijdens lopen, en de vraag hoe het sturingssysteem omspringt met onderling strijdige beperkingen. De transversus abdominis (TA), obliquus internus (OI) en obliquus externus (OE) zijn betrokken bij een aantal verschillende functies: ademhaling, het regelen van de oriëntatie van de romp, en stabilisering van de pelvis en de wervelkolom. De EMG-activiteit van deze spieren werd geanalyseerd, en een model werd gebruikt voor het berekenen van de 3-dimensionale rompmomenten tijdens het lopen op een lopende band met zes verschillende snelheden (1.4-5.4 km/h). Principiële Componenten Analyse liet zien dat de tijdseries van de rompmomenten consistent waren over snelheden en proefpersonen, zij het iets minder in het sagittale vlak. De drie spieren waren elk actief gedurende ≥ 75% van de schredencyclus, hetgeen een stabiliserende functie suggereert. De spier-activiteit werd duidelijk gemoduleerd per fase, met de TA vooral actief tijdens de ipsilaterale zwaaifase, OE contralateraal, en OI grotendeels symmetrisch. Fourier analyse liet vier hoofdfrequenties zien: ademhaling, schredenfrequentie, stapfrequentie, en een tri-fasisch patroon (vermoedelijk een combinatie van schreden- en stapfrequentie). Wanneer de snelheid toenam, bleef de absolute sterkte van alle frequenties gelijk, of nam die toe; de relatieve sterkte van spier-activiteit met het ritme van ademhaling of van de schrede nam af, terwijl de relatieve sterkte toenam van stapgerelateerde activiteiten en van het trifasisch patroon. De effecten van snelheid waren gradueel, en binnen de proefpersonen hadden de lineaire EMG enveloppen een aanzienlijke gezamenlijke variantie (> 70%) over snelheden, hetgeen suggereerde dat de spieren op alle snelheden bij dezelfde taken waren betrokken. Een verdere analyse in het tijdsdomein liet zien dat spieren zowel simultaan als volgtijdelijk betrokken waren bij het uitvoeren van taken. Om het probleem van onderling strijdige beperkingen op te lossen werd co-contractie gebruikt, waarbij antagonisten compenseerden voor de ongewilde neveneffecten van individuele spieren.

In Hoofdstuk 5 wordt een meer gedetailleerde analyse aangeboden van spier-activiteit gedurende de ASLR. In de literatuur wordt dikwijls aangenomen dat stabiliserende activiteit naar voren komt in bilaterale symmetrie. Een gedetailleerde analyse van de ASLR suggereert dat dit niet correct is, en dat de feitelijke activiteit van spieren het gevolg is van het bij elkaar optellen van symmetrische en asymmetrische componenten. In Hoofdstuk 2 werd contralaterale activiteit van een heupstrekker gevonden, met name de m. biceps femoris. Die activiteit drukt de contralaterale hiel naar beneden, en de contralaterale pelvis omhoog, hetgeen een ipsilaterale rotatie van de pelvis impliceert. Deze ipsilaterale pelvis-rotatie werd vooral tegengegaan door ipsilaterale activiteit van de transversus abdominis. Dit is een meer complex voorbeeld van hoe het sturingssysteem het probleem oplost van onderling strijdige beperkingen, namelijk door co-contractie waarbij neveneffecten van de ene spier worden gecompenseerd door activiteit van een andere spier.

Hoofdstuk 6 bevat de algemene discussie van bewegingssturing tijdens de ASLR en tijdens het lopen, vanuit het gezichtspunt van de lumbopelvische stabiliteit. We weten inmiddels dat bilaterale activiteit van de laterale buikspieren kan bijdragen aan lumbopelvische stabilisering; multitasking is een belangrijk verschijnsel; geen enkele spier is beperkt tot slechts één taak; coördinatie pakt het probleem van onderling strijdige beperkingen aan; en het hele systeem heeft een hoge mate van variabiliteit. De belangrijkste conclusie van deze dissertatie is dat toekomstige biomechanische studies van pijn in de bekkengordel zich moeten richten op de 3-dimensionale kinematica en kinetica van (het grootste deel van) het lichaam. Op zichzelf is dit wellicht geen verrassende conclusie, maar tot nu toe zijn er nauwelijks studies gepubliceerd die zich voldoende rekenschap gaven van de relevante 3-dimensionaliteit.

Acknowledgements

Thank you, Dank u, 致谢

It has taken a long journey to change “Dr. Hu” (orthopaedic surgeon) to this other “Dr. Hu” (biomechanist). I still can’t believe that my endeavour has finally turned into a book, but I keep in mind that I would not have survived without the help of many colleagues and friends.

My first acknowledgement goes to the late Prof. P.I.J.M Wuisman. Dear Paul, it is you who offered me the job in this amazing country. We had such a good time together in the hospital. It was marvellous to write a clinical paper with you. I cannot help missing you whenever I see you picture which I have put on my table at home for years.

I am in debt to Onno G Meijer. What an extraordinary guy! You always have brilliant ideas. Yes, you inspired me with many great ideas. You taught me how to write, more important, how to think as a scientist. You tried all the way to extend my contracts, thus allowing me to finish my work here. Onno, you are my friend forever.

I owe my deepest gratitude to Jaap van Dieën. You are my role model! You have such a wide range of knowledge and such a scientific instinct. You are also kind and gracious. I can’t remember how many times you came to my room to tutor me, or helped me to prepare documents I needed. For me, it would have been impossible to become a real researcher without you.

Sjoerd and Yun-Ju, my best friends! It is my honour to meet you two geniuses. Each time we talked, I was inspired. You are always encouraging and helped me to improve my knowledge of Matlab, statistics, EMG, kinematics, kinetics, and modelling. I am glad you two will be my paranymphs.

From the bottom of my heart, I express my great appreciation to Prof. P.W.Hodges. Dear Paul, in Brisbane, in your lab, I started my PHD project for real. I had a good time there, but it was only two weeks. For years, you have given me so many stimulating opinions and suggestions. I am also grateful to my dear friend Henry Tsao and his wife Mei Kuo, treating me with great hospitality during my stay in Brisbane.

Dear Hamid Abbasi and Hamid Fallah, the three of us shared so much happiness, so many difficulties, so much sadness and suffering, for years. Playing ping pong, darts and volleyball was a lot of fun. Both of you are like family to me. Many thanks buddies!

Dear Barend and Margreet, thanks to you, I was lucky enough to stay in this beautiful city without having to change my address even once. Dear Peter Beek, your laughter is impressive, and thank you for your concern for me.

Wu WenHua, you went here before me, and it is a great pleasure to cooperate with you now. Thank you so much. Dear Rob and Prabath, our research covered many different areas, and would have been impossible without your help. Many thanks.

Xia Chun, you are one of the most unusual persons in the world, and it is an honour to work with you. Of course, I missed you and the group in Xiamen, and it is good to come back now.

Dear Hans, Leon and Bert, thank you for your very professional help with the equipment. Dear Andreas, DirkJan, Claudine, Idsart, Maarten, Huub, Erwin, and Jo, it was always a pleasure to have the chance to go to you and ask for advice. You are good teachers. Dear Mark and Ilse, thank you for your technical assistance, and Mark, thank you for lending your body to my first fine wire measurements. Dear Rita Oliveira, Rita Sousa and the other subjects, thank you for joining my experiments. I thank all my previous and present colleagues for creating an excellent studying environment at the FBW. In particular I am grateful to Nienke and Menno for their help.

Despite the occupation of academic work at the FBW, my outside life was far from boring. A “thank you” to my roommates: Lu Zufu, Huang Chunling, Wang Wenlong and Li Xiaolan. We underwent a lot of ups and downs together, and these moments are carved in my heart. I had such a good time with my friends Guo Baoluo, Wu Mengxiao, Liu Yi, Wang Yanjing, Chai Xiangfei, Yu Yong, Zheng Bin, Sui Suli, Huang Zhisheng, Deng Dongmei, Zhang Yanru and Shan Ruizhen. Thank you all!

Sports occupied most of my spare time. I am grateful to my badminton team-mates: Zhuo Qiquan, Jin Kaikai, Fang Yue, Ken, Tang Miao, Liu Xiaolong, Chen Tiexi, Jimmy, Joana, Puikwun Wong, Zhang Jinhua, Xiao zhenxia, Xu Chenjian and Zhou Tian. Peter Huijing, Arjen van Rijn and Yasuo Kawakami, the fun is not only in the tennis court, but also in the drink and talk after exertion.

It is my honour to thank the Lo family (Jane Lo, Henry Lo, Stewen Lo), and the Yang family (Huagen Yang and his daughter Josje) for many joyful conversations and dinner treats. I am heartily thankful to the Chinese Student Association, which allowed me to make many Chinese friends, and we could help each other. Dear Mr. Luo Ping, you always leave your door open; Zhou Cheng, Ding Ning, Duan Xuexin, Maoziqian, Zhang Hetian, Li Yunlei, Zhang Zhen, Rao Xiangyu, Ninghu Ning, Liu Yanchao, Guo Hua and the other colleagues from ACSSNL, thank you all for the cooperation and creative ideas.

It is impossible to list all my friends in this limited space, but I still like to thank Wu Ming, Liu Fenrong, Liu Yang, Zhu Jiansong, Yang Guangyuan, Luo Xi, Zhang Yue, Shan Lin, Qi Xinrui, Yang Ning, Gui Yanwen, Li Jiayang, Liu Fangbin, Huang Zhenxing and Xiedan, Zhou Jia and Wang Hongliang, Wu Jun, Wang Ting, Yuan Taoying, Wang Huajun, Song Shixin, Huang Xuelian, Liu Rui, Wang hang, Ren Shanshan, Angelino, Frederic and the others, who helped me in one way or the other.

Lastly, I am indebted to my family for unselfish love, support and understanding, Mom, I am going back to take care of you!

Curriculum Vitae

Curriculum Vitae

Personal Information

Name	Hai Hu (**Hu** is the last name)	**Gender**	Male
Date of Birth	1 Feb. 1971	**Nationality**	Chinese
Profession	Orthorpaedic Surgeon & PhD of biomechanics		
Contact	E-Mail: xmhuhai@gmail.com		

Working Experience

<table>
<tr><th>Employed by</th><th>Time</th><th>Profession</th><th>Main skills</th></tr>
<tr><td>Hunan Medical University</td><td>1993 - 1995</td><td>Resident surgeon and master student</td><td rowspan="6">• Emergency treatment of Trauma
• Replantation of amputated fingers and limbs
• Replantation of flaps
• Anastomosis of injuried arteries and nerves
• Arthroplasty and hemiarthroplasty
• Fracture reduction and fixation in limb and spine (Surgery & Chinese traditional conservative treatment);
• Rehabilitation for chronic musculoskeletal diseases
• Interventional treatment and biopsy via ultrasound or CT, etc.</td></tr>
<tr><td rowspan="4">Zhongshan Hospital, Xiamen University</td><td>1995 - 1996</td><td>Resident surgeon:
surgery and trauma surgery</td></tr>
<tr><td>1996 - 1998</td><td>Resident orthopaedic surgeon:
hand surgery and trauma surgery</td></tr>
<tr><td>1998-2004</td><td>Attending orthopaedic surgeon and docent:
microsurgery, arthro-surgery and trauma surgery</td></tr>
<tr><td>2004 - 2005</td><td>associate professor:
microsurgery, arthro-surgery and trauma surgery</td></tr>
<tr><td>Shanghai No.6 the People Hospital</td><td>1999 - 2000</td><td>Orthopaedic surgeon in an advanced program:
microsurgery, arthro-surgery and trauma surgery</td></tr>
<tr><td>Faculty of Human Movement Science, Vrije Universiteit, Amsterdam and Department of orthopaedics, VU medical center</td><td>2005 - 2011</td><td>PhD student of biomechanics
Promotor:
Prof. Jaap van Dieën
(+31 205988501);
late Prof. MD. P.I. Wuisman;
Prof. MD. B.J. van Royen;
Dr. Onno G. Meijer</td><td>• Kinematics, Kinetics, EMG (fine wire EMG)
• Ultrasound guided intervention
• Matlab programming
• Image Processing
• Signal Processing</td></tr>
</table>

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- ***Hu H**,* Meijer OG, Van Dieën JH, Hodges PW, Bruijn SM, Strijers RL, Nanayakkara PW, van Royen BJ, Wu WH, Xia C. Symmetry and asymmetry of abdominal wall muscle activity during the Active Straight Leg Raise (ASLR). **Manual Therapy. 2011**, submitted.
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Conferences

- **The 2nd International Fascia Research Congress** (26-31 October 2009, Amsterdam, Netherlands, participant);
- **The XVIII Congress of International Society of Electrophysiology and Kinesiology (ISEK) 2010** (16-19 June 2010, Aalborg, Danmark, poster);
- **The 7th Interdisciplinary World Congress on Low Back & Pelvic Pain 2010** (9-12 November 2010, Los Angeles, USA, poster & presentation, represented by Jaap van Dieen);
- **ISB 2011** (3-7 July 2011, Brussels, Belgium, presentation).